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Numerical Models of Blunt Dissection and Brain

Retraction for Neurosurgery Simulations

(脳神経外科手術シミュレーションのための
鈍的剥離と脳圧排数值モデル)

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Numerical Models of Blunt Dissection and Brain Retraction for Neurosurgery Simulations*

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Abstract

In the brain, parts that control important functions, such as language and motion functions are dense. In neurosurgeries, accidentally damaging the surrounding area of a lesion can result in serious after-effects. Therefore, extreme caution is necessary during neurosurgeries. A neurosurgery simulator is expected as a training method to train novice neurosurgeons in basic surgical skills. Neurosurgery simulators are expected not only for training novice surgeons but also for helping experienced surgeons in preoperation planning and operative procedure confirmation. Several neurosurgery simulators have been developed. In such a haptic surgery simulator, the deformation of the brain tissue is displayed on the display, and the reaction force is presented to the operator through the haptic device.

Basic skills of neurosurgery include sharp dissection, blunt dissection, and brain retraction (to press the surrounding area to secure the path to the lesion). Among them, experiences are especially necessary for blunt dissection and brain retraction. Therefore, development of a neurosurgery simulator that can accurately simulate these two skills is expected.

The knowledge of situations that can cause damage and fracture of brain tissue is very important for the surgeon during a surgery. Macroscopic fracture, such as blunt dissection due to aspiration tube, as well as microscopic damage due to brain retraction can cause damage to the cranial nerve. Therefore, for neurosurgery simulation, it is necessary to construct stress relaxation and damage fracture models that can accurately calculate the stress and reaction force of brain tissue.

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In this study, a finite element method is used as a numerical calculation method to construct a simulation, with a dynamic motion equation. The brain tissue is modeled as a homogeneous, isotropic, linear material.

Most of the previous studies about damage and fracture of brain tissues were conducted with high strain rate for traumatic brain injury. Strain rate dependence exists in the mechanical properties of brain tissues. Therefore, experimental results and numerical models of traumatic brain injury cannot be applied to neurosurgery simulations, which are usually performed at a low strain rate. In this study, we focus on damage of brain tissues at a low strain rate. Tensile tests were conducted using porcine brain tissues, and the mechanical properties of damage of brain tissues were measured.

As the existing damage fracture models cannot reproduce the mechanical characteristics of damaged fracture of brain tissues, new damage fracture models that can reproduce damage of brain tissue are proposed in the present study. The implementation of new models is easy, with low computation cost. In addition, parameters necessary for the new models were identified based on experimental data. Tensile simulations were conducted under the same conditions as the experiments, and it was confirmed that the damage models accurately reproduce damage of the brain parenchyma.

For the viscoelastic model of the brain, many studies have used the generalized Maxwell model so far. However, in the original generalized Maxwell model, the inertial force due to mass of the brain is not considered. When the simulation is analyzed using the finite element method, the stiffness matrix may become a singular matrix when the brain tissue is cut or peeled off. In the deformation calculation of the simulation model, the inverse of the stiffness matrix is usually used. As such, numerical calculation becomes unstable when the stiffness matrix is singular. In this research, we propose a model that stabilizes numerical calculation even if the stiffness matrix becomes singular by adding an inertia term to the generalized Maxwell model. In addition, experiments were conducted using columnar specimens of brain tissues, and parameters were identified. Experimental results were reproduced through a simulation using the proposed method, and the usefulness of the proposed stress relaxation model was verified.

In mechanical experiments using animal brain, most of the previous researchers had cut out a part of the brain for experimentation. However, in such experiments, stress generation depending on the structure of the whole brain could not be measured correctly. Thus in this study, retraction experiments at longitudinal fissures of cerebrum

were conducted using a whole porcine brain, and the reaction force was measured. It was confirmed that the parameters of the cylindrical specimen cannot be used for a whole brain model. Therefore, the parameters of the whole brain model were identified. In addition, simulations were performed using a finite element model of a porcine brain, and the stress characteristics of experiments were reproduced.

The goal of future studies will be the introduction of damage fracture and stress relaxation models into the neurosurgery simulation.

Keywords: neurosurgery simulation, FEM, damage fracture model, stress relaxation model

Contents

1. Introduction	1
1.1 Background	1
1.1.1 Neurosurgery simulations	1
1.1.2 Basic skills of nuerosurgery	3
1.2 Previous Works	5
1.2.1 Developed neurosurgery simulations	5
1.2.2 Damage fracture mechanics and numerical models of brain tis- sue for blunt dissection	7
1.2.3 Stress relaxation mechanics and numerical models of brain tis- sue for brain retraction	9
1.2.4 Biomechanical properties of brain tissue	10
1.3 Research Purpose and Contribution	10
1.3.1 Research purpose	10
1.3.2 Contributions	11
1.4 Outline of the Thesis	12
1.5 Nomenclature	13
2. Finite Element Model of Brain Tissue	17
2.1 Introduction	17
2.2 Geometric Model	18
2.3 Finite Element Method	21
2.3.1 Basic concept of the FEM	21
2.3.2 Dynamic motion equation of FEM	27
2.3.3 Calculation of deformation and force in the dynamic FEM	29
2.4 Measurement of Mechanical Properties [64]	32
2.4.1 Specimen and equipment	33
2.4.2 Measurement of density	34

2.4.3	Measurement of Poisson's ratio	35
2.5	Summary	37
3.	Damage Fracture Model	39
3.1	Introduction	39
3.2	Tensile Tests of Porcine Brain Parenchyma	40
3.2.1	Specimen preparation and equipment	40
3.2.2	Results of tensile tests	41
3.2.3	Differences between brain and other soft biological tissues	42
3.3	Previous Studies on Fracture and Damage Numerical Model	46
3.3.1	Conventional fracture models for biotissues	47
3.3.2	Conventional damage models	49
3.4	Damage Fracture Model A [79]	55
3.4.1	Damage fracture model A	56
3.4.2	Parameter identification for damage fracture model A	58
3.5	Damage Fracture Model B [82]	60
3.5.1	Damage fracture model B	61
3.5.2	Parameter identification for damage fracture model B	61
3.6	Tensile Simulation	63
3.7	Summary	70
4.	Stress Relaxation Model	71
4.1	Introduction	71
4.2	Previous Studies on Viscoelasticity Model	72
4.3	Numerical model of viscoelasticity with an inertial term [85]	75
4.4	Stress Relaxation Tests [85]	76
4.4.1	Stress relaxation tests	76
4.4.2	Data analysis	79
4.5	Stress Relaxation Simulation [85]	79
4.5.1	Simulation scenario	79
4.5.2	Parameter identification	80
4.5.3	Stress relaxation simulation	82
4.6	Retraction Experiments at Longitudinal Fissures of Cerebrum	87

4.7	Brain Retraction Simulation	91
4.7.1	Simulation scenario	91
4.7.2	Parameter identification	93
4.7.3	Simulation results	94
4.8	Summary	94
5.	Conclusion	97
5.1	Summary	97
5.2	Future Work	98
	Acknowledgements	99
	References	101
	List of Publications	111

List of Figures

1.1	Overview of a neurosurgery simulator [1, 2].	2
1.2	Basic skills of neurosurgery	4
1.3	Technique of arachnoid mater dissection using scissors [9].	5
1.4	NeuroTouch and ImmersiveTouch	7
2.1	Human brain model [55].	20
2.2	Porcine brain model [56].	20
2.3	Example of simple shape model.	20
2.4	Basic concept of the finite element method.	22
2.5	The concept of the displacement field in a primary tetrahedral element.	23
2.6	Definition of known and unknown amount by the nodal condition. . . .	30
2.7	Process of calculation of the dynamic motion equation.	32
2.8	Equipment of measurement system	33
2.9	Example of specimens	33
2.10	Measurement results of density and linear approximation.	35
2.11	Equipment for measuring Poisson's ratio.	36
2.12	Illustrator of specimen for Poisson's ratio measurment.	37
2.13	Example of x -axis strain measurement results.	38
2.14	Measurement results of Poisson's ratio and linear approximation. . . .	38
3.1	Overview of the tensile test system.	41
3.2	Specimen illustrator of tensile tests.	42
3.3	Sequences of a tensile test at a velocity of 0.1 mm/s.	43
3.4	Stress—strain curves of brain tissue in tensile tests.	44
3.5	Comparison between fitted curves for data at the low tensile velocity (0.1 mm/s) and the high tensile velocity (1.0 mm/s).	45

3.6	Typical stress—strain curve for brittle fracture seen in tensile tests of skin, liver, and muscle tissues.	45
3.7	Shape and dimensions of the brain tissue model used in the simulation.	47
3.8	Stress—strain curves given by the removing element model.	48
3.9	Stress—strain curves given by the XFEM analysis using ABAQUS. . .	49
3.10	Schematic illustration of continuum damage mechanics.	50
3.11	Stress—strain curves given by the Gurson—Tvergaard (GT) model. . .	50
3.12	A stress—strain curve of a tensile simulation with the simplified Lemaitre damage model.	54
3.13	Schematic illustration of mechanics of damage model.	55
3.14	Step1	56
3.15	Step2	57
3.16	Step3	57
3.17	Image of optimization algorithm.	58
3.18	Identification process.	59
3.19	Stress—strain curve obtained through a dynamic simulation with identified parameters and damage fracture model A, and the curve obtained by fitting to experimental results with tensile velocity 0.1 mm/s.	60
3.20	Stress—strain curve obtained through dynamic simulation using the identified parameters.	62
3.21	Flow chart of the numerical calculation progress with damage fracture model	66
3.22	Sequences of the simulation at a tensile velocity of 0.1 mm/s with damage fracture models A and B.	67
3.23	Stress—strain curves of experiments and the simulation using proposed damage model A.	68
3.24	Stress—strain curves of experiments and the simulation using proposed damage model B.	69
4.1	Schematic of Maxwell and generalized Maxwell models	72
4.2	Measurement equipment of stress relaxation tests with porcine brain parenchyma	76

4.3	Example (specimen No.6) of stress relaxation test results at a strain rate of 0.01 (s^{-1}).	77
4.4	Example (specimen No.6) of $E(t)$ calculated by Fig. 4.3at strains of 0.1, 0.2, and 0.3.	77
4.5	Three experimental results of tensile and stress relaxation tests with a strain rate of 0.01 (s^{-1}) from strain = 0 to strain = 0.1.	78
4.6	Comparison of the experimental results with strain rates of 0.01 (s^{-1}) and 0.1 (s^{-1}).	78
4.7	Fitted curves ($N=1 - 5$) of the average stress–time curve at a strain rate of 0.01 (s^{-1}).	81
4.8	Strain provided to the simulation model and the target of stress in optimization problem (moving average in Fig. 4.5)	82
4.9	Stress–time curve of stress relaxation simulation.	85
4.10	Stress relaxation simulation with identified parameters.	86
4.11	Changing the scale of time in Fig. 4.9(b) to long span.	86
4.12	Experimental equipment for splitting brain fissure.	87
4.13	Retraction force results of experiments.	89
4.14	Sequences of an experiment (specimen No.11). The retraction velocity is 1 (mm/s)	90
4.15	Comparison of average of experimental results of force with retraction velocity of 1 (mm/s) and 10 (mm/s)	90
4.16	FE model of porcine brain (transformed from digital atlas [87, 56]).	91
4.17	Simulation conditions and identified parameters	95
4.18	Retraction simulation	95
4.19	Retraction simulation result and experimental data.	96

List of Tables

1.1	Some important characters	14
1.2	Some important symbols	15
1.3	Some important abbreviations	16
2.1	Sizes of specimens for density measurment.	34
2.2	Measurment results of Poisson's ratio.	37
3.1	Sizes of specimens of tensile tests.	42
3.2	Mechanical properties and damage model parameters of conventional fracture or damage models.	46
3.3	Characters for simplified Lemaitre damage model	52
3.4	Mechanical properties and parameters of proposed damage fracture models A and B for tensile simulations.	65
4.1	Accuracy of fitted curves.	81
4.2	Mechanical properties and parameters of generalized Maxwell model for stress relaxation simulation	83
4.3	Size of the FE model of porcine brain.	91

Chapter 1. Introduction

1.1 Background

1.1.1 Neurosurgery simulations

Brain is the most delicate and complex organ in the human body. It serves as the center of the nervous system and controls many senses such as vision, memory, and motion. Even slight mistakes in neurosurgery could inflict permanent damage, such as spasm, torpor, and memory disorder, on a patient. To perform a neurosurgery successfully, neurosurgeons should memorize amounts of knowledge and practice a lot. Therefore, training and education is very important for neurosurgeons.

Techniques inheritance issues exist in various fields, and the medical field is no exception. Especially in neurosurgeries, training and education of surgical skills are discussed as a subject. To acquire skills, neurosurgeons first practice using animals and plastic models as off-the-job training. Ultimately on-the-job training (i.e. surgical experience) is necessary. However, it is difficult to give surgical opportunities to novices in the current severe social trend. Moreover, neurosurgical institutions and specialists are increasing. Furthermore, depending on symptoms, radiation therapy (e.g., gamma knife and radio wave cautery), chemotherapy with anticancer drugs and intravascular treatment are in widespread use. Therefore, the number of surgical cases experienced by each neurosurgeon is decreasing. If the inheritance relies on the job training, it might become impossible because the number of experienced cases is limited. Therefore, it is necessary to increase the proportion of off-the-job training.

Conventional methods of off-the-job training include the use of animals, plastic models, and cadavers. Experimental animals are usually pigs, dogs, or rats. Sometimes it is difficult to prepare an appropriate animal model depending on the target case. Moreover, there is a limit to use animals due to the surge of animal rights activists. The plastic model is a human body model called a surgical trainer. Its structure and texture

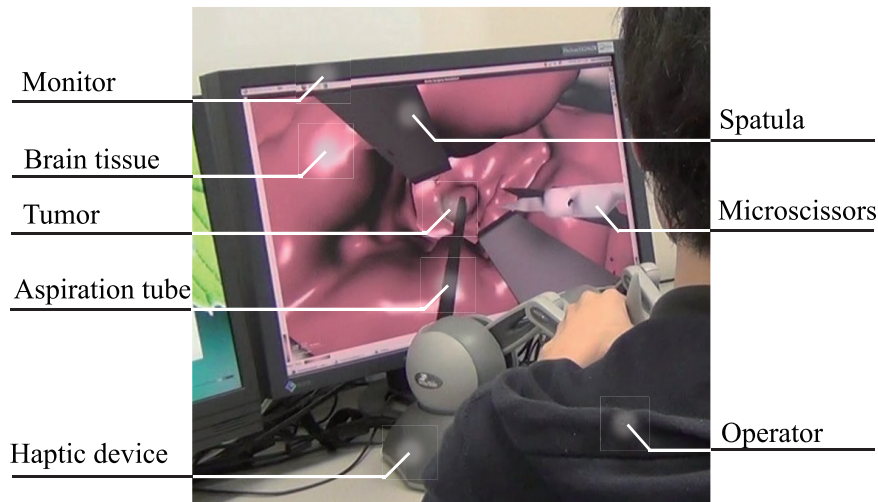


Figure 1.1: Overview of a neurosurgery simulator [1, 2].

is close to that of a living body. However, in neurosurgery, it is difficult to create a model because the brain structure is complicated, and biological reactions cannot be simulated. Cadavers are scarce and biological reactions cannot be simulated. Therefore, animals, models, and cadavers cannot be used to create exactly the same situation of neurosurgeries even if similar situations can be created.

Therefore, conventional training methods cannot meet training requirements of neurosurgeons. With advances in technology, virtual reality (VR) technology plays an important role in the medical field. Surgical simulations are increasingly recognized by surgeons. The surgical simulator is a combination of VR and computer technologies, and can thus create a virtual environment. Surgeons can immerse themselves in virtual environment and experience various situations of an actual surgery.

Figure 1.1 shows an example of neurosurgery simulators [1, 2]. A neurosurgery simulator is generally equipped with haptic devices and video display monitors. Operators feel virtual feedback force via the haptic devices and watch the model image displayed on the monitor.

The virtual surgical system is flexible. In particular, as real patient models reconstructed from magnetic resonance imaging (MRI) and computed tomography (CT) images can be used, it is possible to create an environment similar to that of an actual surgery compared to conventional training methods. The advantages of a surgical simulator using the virtual reality technology are summarized as follows.

(i) **Training basic surgical techniques of novice surgeons**

The surgical simulator can be used by the user to repeatedly train in a surgical procedure. It helps improve surgical skills.

(ii) **Establishing an expert system**

Surgical experience of experts can be collected and analyzed to establish an expert system. Novice surgeons can then study the skills. Moreover, the skills of novice surgeons can be quantitated by specific standard. In addition, the training cost and time can be reduced using the expert system.

(iii) **Preoperative planning**

Precise preoperative planning and optimal procedures can be carefully planned for each patient case. In neurosurgical surgeries, rare cases often occur. Practice can be tested before an actual surgery and the results can be observed. It is possible to quickly and accurately perform surgeries of rare cases even for the first time. By assembling surgical procedures with surgical simulation before actual surgery, the effect of dramatically improving the safety of the neurosurgeries can be expected.

Seymour *et al.* reported that speed of surgical training by using a surgical simulator increased by 30% and the number of mistakes in the operation decreased to 1/6 [3]. The effectiveness of the neurosurgery simulator has also been demonstrated [4].

Consequently, the surgical simulator can reduce the risk of surgery by novice surgeons, improve the success rate of experts, and indirectly increase the quality of life of patients. Therefore, development of a neurosurgery simulation system is very important.

1.1.2 Basic skills of nuerosurgery

The basic skills necessary for neurosurgery include the ability to use a spatula, an aspiration tube, and micro-scissors (Fig. 1.2). A spatula is used to separate brain tissues for lesion exposure and operative field securing. An aspiration tube is used to suction cerebrospinal fluid, perfusion water, and blood in the surgical field. It is also an important retractor. In neurosurgeries, brain surface and blood vessels are towed by aspiration tube to assist the dissection of the arachnoid membrane. In addition, soft tumors can be removed using an aspiration tube. Such separation of tissues along natural lines of

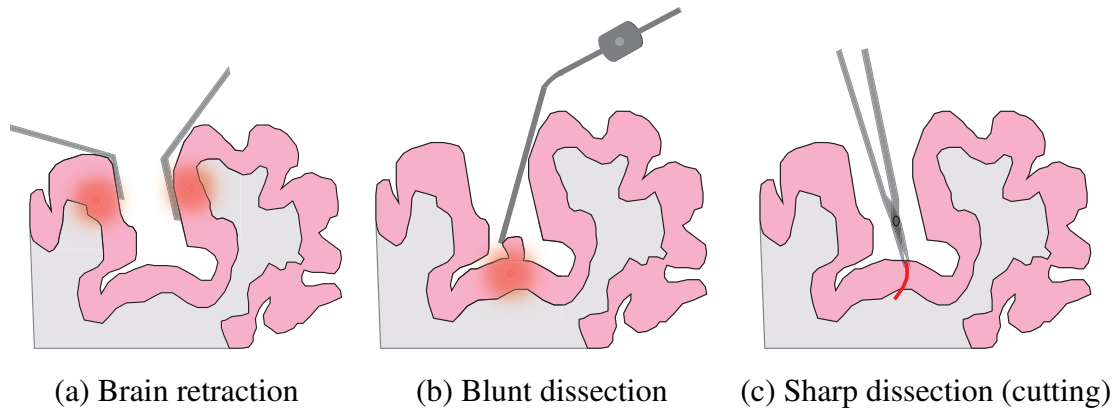


Figure 1.2: Basic skills of neurosurgery

cleavage with blunt instruments is called blunt dissection. Micro scissors are used to perform sharp dissection (also referred to as cutting).

Inappropriate surgical operation easily causes damage to the brain. There are two types of damages: direct sharp damage and indirect blunt damage.

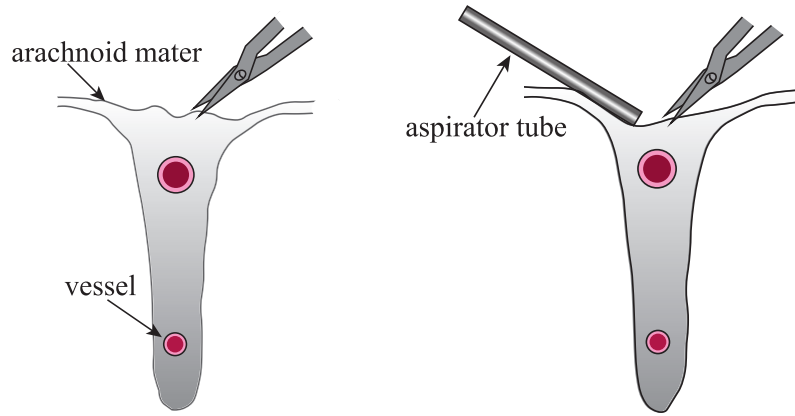
(i) Direct sharp damage

A direct sharp damage is caused by knife or scissors. As shown in Fig. 1.3 (a), when cutting arachnoid mater by using scissors, if the arachnoid mater is slacken, blood vessels and brain parenchyma directly beneath might be damaged. Therefore, one technique is to provide local tension to the brain surface near the side of the lesion to be cut. The tension is given by using an aspiration tube or tweezers, as shown in Fig. 1.3 (b). The tension makes it easier to make a cut. However, even a slightly stronger tension on the brain surface will cause indirect blunt damage, as described later.

(ii) Indirect blunt damage

When excessive tension from the aspiration tube is added to the brain surface, a healthy part is torn off from the brain surface, causing blunt damage. In addition, excessive stress due to brain retraction may cause brain injury such as contusion, infarction, and edema [5, 6, 7, 8].

As shown in Fig. 1.2, basic skills of neurosurgeries can be classified into three types: retraction, blunt dissection, and sharp dissection. As aforementioned, the common difficulty of these skills is the possibility of damaging brain tissue when an operator applies



(a) Arachnoid mater is slacken (b) Arachnoid mater is stretched

Figure 1.3: Technique of arachnoid mater dissection using scissors [9].

excessive force to the brain. Especially, avoiding blunt damage is difficult because it relies on the sense of force or stress. To acquire this technique, considerable amount of time and experience is required. Therefore, to train appropriate techniques of retraction and blunt dissection in brain surgery simulation, accurate presentation of stress distribution in brain tissues and damage fracture model are required.

1.2 Previous Works

To achieve an accurate presentation of stress distribution in brain tissues and damage fracture model, several difficulties must be overcome. The related research on the issues stated in Section 1.1 are as follows.

1.2.1 Developed neurosurgery simulations

In general, there are three important integrants in a surgical simulator.

(i) Reaction force presentation

Surgeons acquire various information from a reaction force, such as specifying the lesion part from the difference in hardness between the normal living tissue and the affected part. Especially in neurosurgery, surgeons often rely on reaction force. Therefore, the accurate presentation of reaction force in reproducing virtual environment is an indispensable integrant.

(ii) Visual presentation

Humans depend more heavily on vision than other senses. Especially in neurosurgeries, reaction force is small, and complementation of information by visual sense is important for operations. Therefore, it is necessary to present accurate tissue deformation.

(iii) Real time

The simulator can accurately calculate the reaction force and deformation. If the simulator cannot present this information at an appropriate timing, it becomes wrong information rather than meaningless. To present the reaction force and deformation with appropriate timing without discomfort, it is necessary to realize a simulation at a speed, which can be recognized by humans. The real time is considered as 30 (ms). Therefore, the sampling time of the computational calculation should be less than 30 (ms) securing stability.

A number of simulators, such as those for endoscopic surgery and pericardial manipulation have been developed for various organs[10, 11]. Some of them have been commercialized successfully, such as the LapVRTM Surgical Simulator (CAE Healthcare Inc.), and the dV-Trainer® (Mimic Technologies Inc.). The validity of these simulators has been confirmed and reported [12, 13].

Compared with laparoscopic surgery simulators, the number of neurosurgery simulators has been limited because neurosurgery simulators need more accurate calculation and the brain is complex. For instance, in usual cancer removal surgery, the surrounding tissues of the lesion are removed together so as not to leave cancer cells. In neurosurgeries, if the surrounding tissues of the lesion are removed, there is a risk that the language or motion function may be lost. Therefore, it is necessary to carefully remove the lesion part as much as possible while leaving the surrounding tissue as much as possible.

So far a few attempts of neurosurgery simulators have been made. Luciano *et al.* developed a neurosurgery simulator named *ImmersiveTouch* (Fig. 1.4 (b))[14], in which only the surface model is used. It can present the operator with resistance when inserting a catheter. However, contact reaction force is not calculated. Delorme *et al.* reported a neurosurgery simulator *NeuroTouch* (Fig. 1.4 (a)) and accomplished a tumor-debulking task [15]. In *NeuroTouch*, only a limited area around the lesion area is modeled. Some



(a) NeuroVR (formerly NeuroTouch) [20]



(b) ImmersiveTouch [21]

Figure 1.4: NeuroTouch and ImmersiveTouch

previous studies presented simulations of cerebral aneurysm clipping [16, 17]. Although usefulness of neurosurgery simulators such as *ImmersiveTouch* and *NeuroTouch* have been confirmed [18, 19], most of them deal with single scenarios in neurosurgery. To realize a neurosurgery simulation that can deal with a series of surgical operations using a whole brain, a neurosurgery simulator was developed in our laboratory[1, 2].

1.2.2 Damage fracture mechanics and numerical models of brain tissue for blunt dissection

The sharp dissection (cutting) of biotissues has been studied and some effective methods have been developed to express cut wounds. The common methods include element subdivision [22, 23] and node snapping [24, 25]. However, in the element subdivision method, the total stiffness matrix requires recalculation. With the rapidly increasing number of elements, this method is computationally expensive. In the node-

snapping method, ill-shaped elements are easily generated causing computational instability. Therefore, both element subdivision and node snapping are not suitable for real-time neurosurgery simulation.

Further, there are few studies regarding blunt dissection of biotissues. Delorme *et al.* and Mora *et al.* presented tissue separation by using a volume-sculpting technique [26, 15]. In [15], the stiffness of tissues was reflected. In [26], the contact force was presented. However, when fractures occur, reaction forces were not discussed.

In these studies, the fracture mechanics were not discussed. A field of research, fracture mechanics, characterizes fracture mechanics of materials, that is crack growth. Fracture mechanics focuses on crack growth and reproduces the brittle fractures accurately. However, soft biotissues, such as brain tissue, present ductile fractures, which cannot be satisfactorily reproduced through conventional fracture mechanics.

In recent years, the field of research, continuum damage mechanics (CDM) are conducted to express ductile fractures [27]. Some damage models have been proposed including the Lemaitre model [28] and Gurson model [29]. These models are used for metals and require the calculation of plastic strain; the damage progresses only in the plastic region. The plastic strain is usually calculated through iterative calculation, which is not suitable for real-time calculation. Moreover, it is difficult to define a yield function that specifies elastic and plastic regions for biological tissues such as brain tissue. Therefore, conventional damage models cannot reproduce the fractures of brain tissue.

To represent accurate damage of brain tissue in a neurosurgery simulation, the damage mechanics of brain tissue should be clarified through experiments. Many studies have been conducted on the mechanics of brain tissue [30, 31]. The experiments were performed using tensile tests [32], compression tests [33, 34, 35], and shear tests [36]. However, most of them focused on the deformation of brain tissue before failure. Although fracture mechanics of other biotissues, such as skin [37], liver [38], and muscle [39], have been investigated, fracture and damage mechanics of brain tissue have not been investigated thoroughly.

Some earlier studies about damage and fracture mechanics of brain tissue target traumatic brain injury (TBI), which is caused by the impact of high strain rate [40, 34]. For example, in [34], the strain rate is $1000 - 3000 \text{ (s}^{-1}\text{)}$. However, there is a strain rate dependency in mechanical behavior of brain tissue. Neurosurgeries are usually conducted

under low strain rate. Therefore, experimental results and numerical models of TBI are not applicable for neurosurgery simulators. Furthermore, in the analysis of TBI, the University of Strasbourg developed the finite element human head model (FEHM) [41]. The model is analyzed using a nonlinear finite element (FE) analysis application called LS-DYNA® [42]. The analysis is offline, and is conducted over several hours [43]. In contrast, the surgical simulation is real time. Therefore, their purposes are completely different.

These aforementioned studies show that the damage mechanics of brain tissues should be clarified through experiments and a damage model should be developed for brain tissues.

1.2.3 Stress relaxation mechanics and numerical models of brain tissue for brain retraction

Several studies have reported that the stress–strain behavior of brain tissues shows time dependency. The test objects include the brain tissue of human [44], swine [45, 46], bovine [47], and rat [48]. To calculate the stress and force correctly, the time dependency of stress–strain behavior of brain tissues should be introduced into neurosurgery simulations. In brain retraction, the strain of brain is constant; this relaxes the stress with time. This phenomenon is called stress relaxation. The Maxwell model has been widely used to reproduce the stress relaxation. For solids, a generalized Maxwell model [49] comprising several sets of Maxwell models is used.

The generalized Maxwell model favorably represents viscoelastic characteristics of biotissues. Several studies have reported the analysis results of brain tissue obtained using the generalized Maxwell model [48, 45, 47]. However, the original generalized Maxwell model does not have an inertial term. Therefore, most of the studies did not consider inertial force caused by mass property of organs. In a neurosurgery simulation, excessive force caused by the movement of surgical instruments must be evaluated. Therefore, an inertial term must be included in the model. Moreover, in the dissection part of a neurosurgery simulation, isolated elements or structurally unstable elements are easily generated in the brain FE model. In the FE method (FEM), displacement of nodes is generally calculated by the inversion of the stiffness matrix. Isolated elements

or structurally unstable elements make the stiffness matrix singular, which could make the computation of the displacement of nodes unstable. By considering an inertial term, even the stiffness matrix becomes singular, and therefore the computation is stable.

Furthermore, in most earlier studies, mechanical characteristics of brain have been tested with a part of a brain [48, 46]. Such tests cannot correctly measure the stress generation that depends on the structure of the entire brain.

From these cited studies, a stress relaxation model should be developed for brain tissues, and stress relaxation mechanics of brain tissues should be clarified through experiments by using an entire brain.

1.2.4 Biomechanical properties of brain tissue

To achieve realistic performance, the biomechanical properties of brain tissues should be set to the neurosurgery simulation. The mechanical properties of brain tissues were first reported by Franke *et al.* [50], and have since been studied for 60 years; thus, a large collection of data is available. However, there is a wide range of measured brain mechanical properties [51, 52] because studies used different testing protocols for different situations. Furthermore, most of these studies put priority on accuracy rather than real-time computation. The biomechanical properties were identified for nonlinear deformation models [35, 53, 33, 54]. Therefore, in this study, the biomechanical properties are identified for each model.

1.3 Research Purpose and Contribution

1.3.1 Research purpose

In this thesis, to accurately simulate blunt dissection and brain retraction of neurosurgery, damage models of brain tissues and a stress relaxation model are proposed. The models are based on the experimental result obtained from tensile and stress relaxation tests using porcine brain tissue.

1.3.2 Contributions

The contributions of this thesis are as follows.

Damage fracture model

1. Most of the previous studies about damage and fracture of brain tissues were conducted with a high strain rate for traumatic brain injury. The mechanical properties of brain tissues are depend on the strain rate. Therefore, experimental results and numerical models of traumatic brain injury cannot be applied to neurosurgery simulations, which are usually performed at a low strain rate. In this study, we focused on the damage of brain tissues at a low strain rate. Tensile tests were conducted using porcine brain tissues, and the mechanical properties of the damage of brain tissues were measured.
2. As the existing damage fracture models cannot reproduce the mechanical characteristics of damaged fracture of brain tissues, new damage fracture models that can reproduce damage of brain tissue are proposed in this study. The implementation of new models is easy, and the computation cost is small. In addition, parameters necessary for the new models were identified based on experimental data. Tensile simulations were conducted under the same conditions as those in the experiments, and the damage models were conformed to accurately reproduce damage of the brain parenchyma.

Stress relaxation model

1. For the viscoelastic model of the brain, many studies have used the generalized Maxwell model so far. However, in the original generalized Maxwell model, inertial force due to brain mass is not considered. When the simulation is analyzed using the FEM, the stiffness matrix may become a singular matrix when the brain tissue is cut or peeled off. In the deformation calculation of the simulation model, the inverse of the stiffness matrix is usually used. In this case, numerical calculation becomes unstable when the stiffness matrix is singular. In this research, we propose a model that stabilizes numerical calculation, even if the stiffness matrix

becomes singular, by adding an inertia term to the generalized Maxwell model. In addition, experiments were conducted using columnar specimens of brain tissues, and parameters were identified. Experimental results were reproduced through simulations using the proposed method, and the usefulness of the proposed stress relaxation model was verified.

2. In mechanical experiments using animal brains, most researches had cut out a part of the brain to do experiments so far. However, in such experiments, stress generation depending on the structure of the whole brain could not be measured correctly. In the present study, retraction experiments at longitudinal fissures of the cerebrum were conducted using a whole porcine brain, and the reaction force was measured. It was confirmed that the parameters of the cylindrical specimen cannot be used for a whole brain model. Therefore, the parameters of the whole brain model were identified. Simulations were performed using a FE model of porcine brain, and the stress characteristics of experiments were reproduced.

1.4 Outline of the Thesis

The thesis consists of five chapters.

Chapter 1 is an introduction.

In Chapter 2, the basic principle of the FEM is described, which is a numerical calculation method to construct simulations. The stiffness matrix for deformation and force calculation of a simulation model is presented. In addition, for the dynamic linear model, the motion equation and the calculation procedure using the FEM are described. Furthermore, the density and Poisson's ratio of brain tissues are measured.

In Chapter 3, two numerical models for blunt dissection, which is one of the basic skills of neurosurgeries, are proposed. First, we present previous researches on damage and fracture of soft body tissues. In addition, the problems of previous research are shown, and the position of this research is explained. Second, two damage models are proposed and formulations of the models are described.

In addition, the proposed numerical models are experimentally evaluated. At first, tensile tests using porcine brain parenchyma are described. Next, a method of identifying parameters of models based on the experimental results is described. Finally,

the experimental results are reproduced through simulations using the damage fracture models and identified parameters. By comparing the simulation and the experimental results, the usefulness of the proposed numerical models is verified.

In Chapter 4, a numerical model for brain retraction, which is one of the basic skills of neurosurgeries, is proposed. First, we introduce previous researches on stress relaxation of brain tissues and their problems, and the position of this research is explained. Second, we propose a stress relaxation model and describe its formulations.

In addition, the proposed numerical model is experimentally evaluated. First, stress relaxation experiments using cylindrical specimens taken from porcine brain parenchyma are described. Next, the parameters of the model are identified. Finally, experimental results are reproduced through simulations by using the stress relaxation model and identified parameters. By comparing the simulation and experimental results, the usefulness of the proposed numerical model is verified.

Furthermore, retraction experiments using pig brain are described, and retraction simulations are conducted.

Chapter 5 summarizes the thesis and suggests future works.

1.5 Nomenclature

In this thesis, specific characters are assigned to mathematical entities as uniformly as possible. Simultaneously, technical terms, signs, and symbols generally accepted in the field are used. Sometimes, it is difficult to satisfy the both objectives. Therefore, in some cases, original characters are adopted. The meaning of some important characters are listed in Table 1.1.

The mathematical meanings of font styles are as follows.

- Italic light-face letters (e.g. E , λ): scalars
- Upright bold-face letters (e.g. $\mathbf{K}$, $\mathbf{u}$, $\boldsymbol{\sigma}$): vectors or matrices

Superscripts, subscripts, and symbols are used as indexes. The meanings of some important indexes are listed in Table 1.2.

Some technical terms are described through abbreviations. Table 1.3 lists some important abbreviations.

Table 1.1: Some important characters

for FEM			
K	total stiffness matrix	E	Young's modulus
C	damping matrix	ρ	density
M	mass matrix	ν	Poisson's ratio
B_e	strain-displacement matrix	V_e	element volume
D_e	elasticity matrix	a, b	Rayleigh damping scalars
K_e	element stiffness matrix	Δt	sampling time
f	force vector	σ	stress tensor
f_{re}	reaction force vector	ϵ	strain tensor
u	displacement vector	$\dot{\mathbf{u}}$	velocity vector
$\ddot{\mathbf{u}}$	acceleration vector		
for damage fracture model			
D	damage variable	σ_t	threshold of damage generation
ω	material integrity	ω_t	threshold of fracture generation
V_{void}	void volume of an element	λ	void volume growth rate coefficient
σ_{Mises}	von Mises stress		
for viscoelastic model (generalized Maxwell model)			
N	number of Maxwell models	η_i	material coefficient of viscosity
E_i	elastic modulus	τ_i	relaxation time
for optimization problem			
x	parameter vector	$\mathcal{F}(\mathbf{x})$	objective function
x₀	initial parameter vector	x_{opt}	optimization result

Table 1.2: Some important symbols

$(\tilde{*})$	effective $(*)$ ($= (1 - D)(*)$)
$(\dot{*})$	material time derivative of $(*)$
$(*)^{\text{trial}}$	trial state of $(*)$
$(*)_{\text{e}}$	an element part of $(*)$
$(*)^{(n)}$	current increment step of $(*)$
$(*)^{(n+1)}$	the next increment step of $(*)^{(n)}$
$\Delta(*)$	increment of $(*)$, $\Delta(*) = (*)^{(n+1)} - (*)^{(n)}$
$(*)^{\text{T}}$	transpose of matrix $(*)$
$(*)^{-1}$	inverse of matrix $(*)$
$(*)_{\text{c}}$	contact nodes part of $(*)$
$(*)_{\text{f}}$	free nodes part of $(*)$

Table 1.3: Some important abbreviations

R^2	Determination coefficient
VR	Virtual Reality
3D	Three-dimensional
FE	Finite Element
FEM	Finite Element Method
MRI	Magnetic Resonance Imaging
CT	Computed Tomography
LM matrix	Lumped Mass matrix
CM matrix	Consistent Mass matrix
TBI	Traumatic Brain Injury
RE method	Removing Element method
XFEM	eXtended Finite Element Method
GT model	Gurson-Tvergaard model
SL model	Simplified Lemaitre Model
CDM	Continuum Damage Mechanics
RVE	Representative Volume Element

Chapter 2. Finite Element Model of Brain Tissue

2.1 Introduction

A continuum analysis is often described by a differential equation using a continuum model. However, when the shape and boundary conditions of structures become complicated, it is difficult to derive the differential equations. Therefore, analysis of such structures becomes impossible. In such a case, continuum, which has infinite freedom degrees, is approximately described using finite freedom degrees. This method is called as discrete analysis. FEM is one of the discrete analysis methods.

In FEM, the object is virtually divided into finite size elements and analyzed as a collection of elements. The method can represent properties such as elastic, plastic, and elastic–plastic. It is commonly used in structural analysis. Recently, the FEM is being widely adopted to solve advanced nonlinear problems, such as plastic processing, vehicle collision, and brain damage mechanism.

Advantages of the FEM are listed as follows.

- Mechanical properties can be described directly, such as Young's modulus and Poisson's ratio.
- Nonlinearity, anisotropy, and boundary conditions can be theoretically modeled.
- Calculation accuracies of deformation and force are high.

The disadvantages of the FEM are that, the stiffness equation that describes the relationship between the deformation and the force of the object model is a large-scale simultaneous linear equation and requires high computational cost.

The procedures of the FEM are as follows.

1. discretization
2. shape function (interpolation function) selection:
3. element matrix derivation
4. entire matrix assembly
5. unknowns derivation

In this chapter, the details of numerical calculation model and method for neurosurgery simulations are presented. The outline of each section is summarized as follows.

Section 2.2 describes the geometric models of brain tissue. In the virtual space, it is necessary to create a geometrical model. In this study, the whole brain models of the human and pig are created from medical images such as MRI. Test specimen models with simple geometric shapes are created manually. In addition, the geometric model is discretized using tetrahedral elements.

In Section 2.3, the FEM is used to describe the relationship between the deformation and force of the object model. The actual brain tissue is known as a nonlinear anisotropic viscoelastic body. To reduce the computational cost, in this study, the model is approximated as a linear isotropic elastic body. In addition, the dynamic motion equation is formulated. An implicit method is used in the dynamic analysis because of its calculation stability.

Section 2.4 presents the measurements of the parameters required for the motion equation described in Section 2.3. The required parameters are density, Poisson's ratio, Young's modulus, and Rayleigh damping coefficients. Density and Poisson's ratio were measured using porcine brains. Young's modulus has large individual differences because of different measurement conditions such as post-mortem time. Moreover, Rayleigh damping coefficients have no physical meaning. Therefore, Rayleigh damping coefficients and Young's modulus were identified in each test later.

2.2 Geometric Model

The actual object is a continuum. To express and calculate the object by using the numerical model, it is divided into elements. By using the geometric model, deformation

of the object can be represented and visual information can be presented to the operator. Moreover, the contact condition among objects and reaction force can be calculated. The procedure of the geometric model is described as follows.

- three-dimensional (3D) geometric model creation
- elements generation
- model data file storing

3D geometric model creation

A 3D geometric model of organs is generally restored from a medical image such as CT or MRI. For example, a multi-modality MRI-based atlas of the brain of a 42-year-old male is available for download [55], and a MRI-based 3D digital atlas of the brain of a female domestic pig is available for download [56]. The atlas data is available under terms of the 3D Slicer [57]. The 3D models in [55] and [56] are shown in Figs. 2.1 and 2.2, respectively. In this study, when the simulation object is an organ, the geometric model is created using atlas data. On the other hand, when the shape of the simulation object is simple, it is created manually by using 3D-CAD (e.g., SOLIDWORKS[58]) . Fig. 2.3 illustrates a simple shaped model.

Elements generation

In this study, the model is divided into tetrahedra. A tetrahedron is a solid with four plane triangular faces. A variety of shapes can be represented using tetrahedra. The software GiD [59] was selected to generate elements. The rich functions of this software include the following points.

- It has a rich element division options.
- It is possible to create a complex 3D shape model.
- It corresponds to a number of file formats of data to read and write.

Examples of human brain, porcine brain, and simple shape model divided into tetrahedra using GiD are shown in Figs. 2.1, 2.2, and 2.3, respectively.

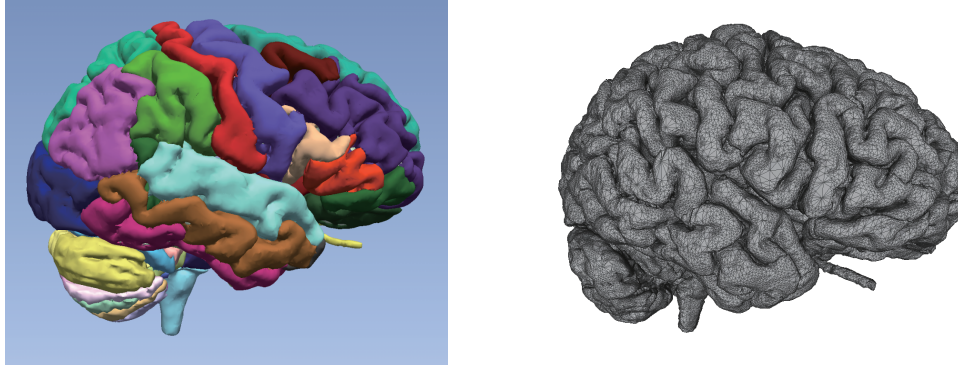


Figure 2.1: Human brain model [55].

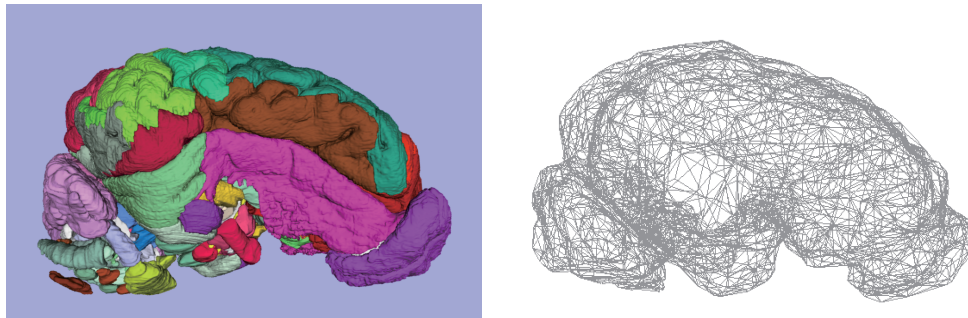


Figure 2.2: Porcine brain model [56].

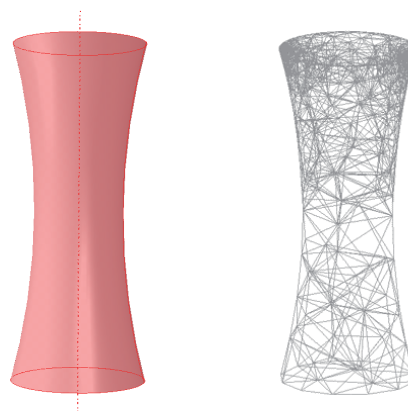


Figure 2.3: Example of simple shape model.

Model data file storing

For efficient calculation, it is necessary to create a data file of the model. In this study, the model is divided into tetrahedral elements each constituting by four nodes. Therefore, the following data information is necessary.

- Node number and the position of the node.
- Tetrahedral element number and number of four nodes

The software GID can be used to write such data into a single msh file.

2.3 Finite Element Method

2.3.1 Basic concept of the FEM

The concept of the FEM is shown as follows

- **FE division of the continuum**

In the FEM, the continuum is divided into FEs as shown in Fig. 2.4. Nodes are located in vertices of elements, which are connected by nodes. The displacement and force of the nodes are unknowns and should be calculated. A shape function is defined for each element. The displacement field in the element is determined from the displacement and shape function of the node.

- **Nodes of elements**

In the FEM, it is assumed that only the nodes can transmit force. To maintain the balance of forces in the object, the nodal force is formulated using a virtual work principle.

- **displacement in element**

The shape function is defined for each element, and the displacement field in the elements is uniquely determined through the displacement of the nodes and the shape function. In this study, tetrahedral elements are assumed as constant strain elements and the shape function is linear. Therefore, displacement and strain in an element are uniform values. In addition, the stress of an element is defined by the displacement field, so that the stress is also a uniform value in an element.

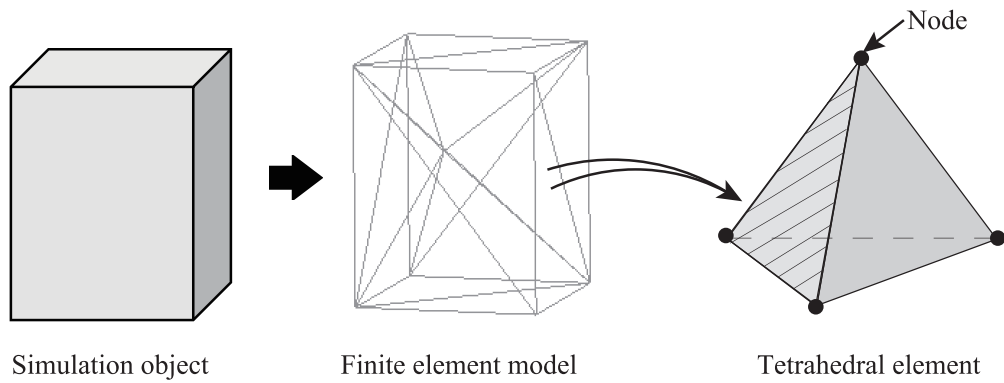


Figure 2.4: Basic concept of the finite element method.

Generally, in simulations, computation failure (or divergence of calculation result) due to discretization error becomes a problem. Moreover, in numerical analysis, this problem can usually be avoided by adjusting the size of step time (sampling time). However, this is impossible in real-time simulation because calculation time is limited in one step time. Typically, the limit is 30 (ms). Thus, in real-time simulation, a calculation technique that emphasizes the stability of behavior while sacrificing accuracy to some extent is indispensable. Almost all soft body tissues are now considered to be nonlinear, anisotropic, and viscoelastic in nature. In this study, to realize a real-time haptic simulation, accuracy is sacrificed to some extent as follows.

- **Inhomogeneous $\rightarrow$ Homogeneous**

Brain tissue is composed of gray matter, white matter, pia mater, arachnoid mater, blood vessel, and so on. In the cerebrum, gray and white matter are primary parts. Gray matter contains numerous cell bodies, and white matter contains mainly myelinated axons. Some researchers have found no difference between the elastic properties of the white matter and gray matter, while others have found differences [60]. In this study, we assumed that white and gray matter have the same elastic properties, and did not separate them when preparing the specimens.

- **Anisotropic $\rightarrow$ Isotropic**

Previous researches reported that the anisotropy of brain tissue is relatively low [61, 62]. In this study, the brain tissue is assumed as an isotropic material.

- **Nonlinear $\rightarrow$ Linear**

Neurosurgery is generally performed in a manner to avoid the application of a

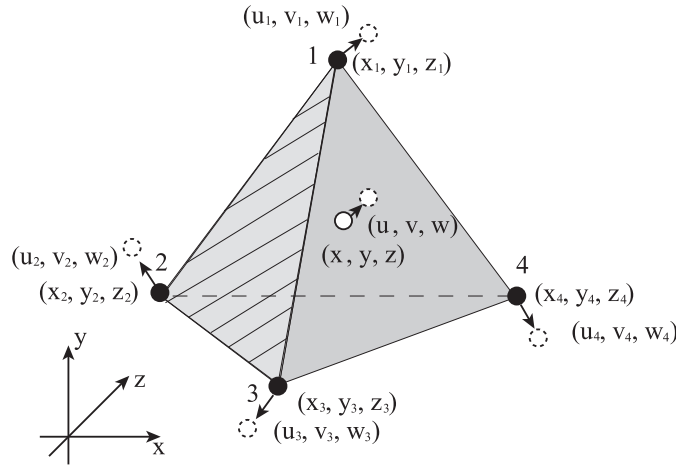


Figure 2.5: The concept of the displacement field in a primary tetrahedral element.

large force on the brain; hence, brain tissues deform linearly in most cases during the surgery.

For all of the aforementioned, brain tissue is assumed as a homogeneous, isotropic, and linear brain tissue model.

In the FEM, deformation and force of a model are calculated using the stiffness matrix: a variable matrix which varies with the shape of the model. Real-time calculation of FEM is difficult because of the stiffness matrix recalculation. However, in this study, stiffness matrix is a constant matrix because the brain tissue is simplified to an isotropic linear model. Therefore, real-time calculation is possible.

In the linear FEM, the deformation and force of an element are described as follows.

Displacement field

First, displacement field is illustrated. Displacement of a point in the primary element is approximated by the interpolating shape function from the node position. In a primary tetrahedral element, the interpolation is obtained from the four nodes. In particular, Fig. 2.5 shows the shape function of the xyz coordinate. In the figure, the coordinates with nodes 1, 2, 3, 4 are $(x_i, y_i, z_i), i = 1, 2, 3, 4$, and the displacement of nodes are $(u_i, v_i, w_i), i = 1, 2, 3, 4$. The optimal shape function that determines displacement

(u, v, w) of a point (x, y, z) in the element is expressed as follows.

$$\begin{aligned} u &= \alpha_1 + \alpha_2 x + \alpha_3 y + \alpha_4 z, \\ v &= \alpha_5 + \alpha_6 x + \alpha_7 y + \alpha_8 z, \\ w &= \alpha_9 + \alpha_{10} x + \alpha_{11} y + \alpha_{12} z, \end{aligned} \quad (2.1)$$

where α_i ($i = 1 - 12$) are unknowns. By substituting (u_i, v_i, w_i) , $i = 1, 2, 3, 4$ into Eq. (2.1), it can be expressed as a matrix as follows:

$$\begin{Bmatrix} u_1 \\ v_1 \\ w_1 \\ u_2 \\ v_2 \\ w_2 \\ u_3 \\ v_3 \\ w_3 \\ u_4 \\ v_4 \\ w_4 \end{Bmatrix} = \begin{bmatrix} 1 & x_1 & y_1 & z_1 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 \\ 0 & 0 & 0 & 0 & 1 & x_1 & y_1 & z_1 & 0 & 0 & 0 & 0 \\ 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 1 & x_1 & y_1 & z_1 \\ 1 & x_2 & y_2 & z_2 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 \\ 0 & 0 & 0 & 0 & 1 & x_2 & y_2 & z_2 & 0 & 0 & 0 & 0 \\ 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 1 & x_2 & y_2 & z_2 \\ 1 & x_3 & y_3 & z_3 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 \\ 0 & 0 & 0 & 0 & 1 & x_3 & y_3 & z_3 & 0 & 0 & 0 & 0 \\ 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 1 & x_3 & y_3 & z_3 \\ 1 & x_4 & y_4 & z_4 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 \\ 0 & 0 & 0 & 0 & 1 & x_4 & y_4 & z_4 & 0 & 0 & 0 & 0 \\ 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 1 & x_4 & y_4 & z_4 \end{bmatrix} \begin{Bmatrix} \alpha_1 \\ \alpha_2 \\ \alpha_3 \\ \alpha_4 \\ \alpha_5 \\ \alpha_6 \\ \alpha_7 \\ \alpha_8 \\ \alpha_9 \\ \alpha_{10} \\ \alpha_{11} \\ \alpha_{12} \end{Bmatrix}. \quad (2.2)$$

$$\mathbf{u}_e = \mathbf{A}\alpha, \quad (2.3)$$

where $\mathbf{A}$ is node displacement coefficient matrix.

Strain–displacement relational expression

Engineering strain $\epsilon = (\epsilon_x, \epsilon_y, \epsilon_z, \gamma_{xy}, \gamma_{yz}, \gamma_{zx})$ is calculated as follows.

$$\begin{aligned} \epsilon_x &= \frac{\partial u}{\partial x}, & \epsilon_y &= \frac{\partial v}{\partial y}, & \epsilon_z &= \frac{\partial w}{\partial z} \\ \gamma_{xy} &= \frac{\partial u}{\partial y} + \frac{\partial v}{\partial x}, & \gamma_{yz} &= \frac{\partial v}{\partial z} + \frac{\partial w}{\partial y}, & \gamma_{zx} &= \frac{\partial w}{\partial x} + \frac{\partial u}{\partial z}. \end{aligned} \quad (2.4)$$

By substituting Eq. (2.4) into Eq. (2.1),

$$\begin{aligned} \epsilon_x &= \alpha_2, & \epsilon_y &= \alpha_7, & \epsilon_z &= \alpha_{12} \\ \gamma_{xy} &= \alpha_3 + \alpha_6, & \gamma_{yz} &= \alpha_8 + \alpha_{11}, & \gamma_{zx} &= \alpha_4 + \alpha_{10}. \end{aligned} \quad (2.5)$$

In addition, by substituting Eq. (2.5) into Eq. (2.2),

$$\begin{Bmatrix} \varepsilon_x \\ \varepsilon_y \\ \varepsilon_z \\ \gamma_{xy} \\ \gamma_{yz} \\ \gamma_{zx} \end{Bmatrix} = \begin{bmatrix} 0 & 1 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 \\ 0 & 0 & 0 & 0 & 0 & 0 & 1 & 0 & 0 & 0 & 0 & 0 \\ 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 1 \\ 0 & 0 & 1 & 0 & 0 & 1 & 0 & 0 & 0 & 0 & 0 & 0 \\ 0 & 0 & 0 & 0 & 0 & 0 & 0 & 1 & 0 & 0 & 1 & 0 \\ 0 & 0 & 0 & 1 & 0 & 0 & 0 & 0 & 0 & 1 & 0 & 0 \end{bmatrix} \begin{Bmatrix} \alpha_1 \\ \alpha_2 \\ \alpha_3 \\ \vdots \\ \alpha_{11} \\ \alpha_{12} \end{Bmatrix} \quad (2.6)$$

Further, by substituting Eq. (2.3) into Eq. (2.6), the relationship between strain and displacement is

$$\varepsilon_e = \mathbf{B}'\alpha = \mathbf{B}'\mathbf{A}^{-1}\mathbf{u}_e = \mathbf{B}_e\mathbf{u}_e, \quad (2.7)$$

where $\mathbf{B}_e$ is the strain–displacement matrix.

Stress–strain relational expression

Based on Hooke's law, in the linear, elastic, isotropic material, the relationship between the strain and stress is expressed as follows.

$$\begin{aligned} \varepsilon_x &= \frac{\sigma_x - \nu(\sigma_y + \sigma_z)}{E}, \quad \varepsilon_y = \frac{\sigma_y - \nu(\sigma_z + \sigma_x)}{E}, \quad \varepsilon_z = \frac{\sigma_z - \nu(\sigma_x + \sigma_y)}{E} \\ \gamma_{xy} &= \frac{\tau_{xy}}{G}, \quad \gamma_{yz} = \frac{\tau_{yz}}{G}, \quad \gamma_{zx} = \frac{\tau_{zx}}{G}, \end{aligned} \quad (2.8)$$

where E , G , and ν are Young's modulus, shearing modulus, and Poisson's ratio, respectively.

Since $E = 2G(\nu + 1)$, Eq. (2.8) is rewritten as

$$\begin{Bmatrix} \sigma_x \\ \sigma_y \\ \sigma_z \\ \tau_{xy} \\ \tau_{yz} \\ \tau_{zx} \end{Bmatrix} = \frac{E(1-\nu)}{(1+\nu)(1-2\nu)} \begin{bmatrix} 1 & \frac{\nu}{1-\nu} & \frac{\nu}{1-\nu} & 0 & 0 & 0 \\ \frac{\nu}{1-\nu} & 1 & \frac{\nu}{1-\nu} & 0 & 0 & 0 \\ \frac{\nu}{1-\nu} & \frac{\nu}{1-\nu} & 1 & 0 & 0 & 0 \\ 0 & 0 & 0 & \frac{1-2\nu}{2(1-\nu)} & 0 & 0 \\ 0 & 0 & 0 & 0 & \frac{1-2\nu}{2(1-\nu)} & 0 \\ 0 & 0 & 0 & 0 & 0 & \frac{1-2\nu}{2(1-\nu)} \end{bmatrix} \begin{Bmatrix} \varepsilon_x \\ \varepsilon_y \\ \varepsilon_z \\ \gamma_{xy} \\ \gamma_{yz} \\ \gamma_{zx} \end{Bmatrix} \quad (2.9)$$

$$\sigma_e = \mathbf{D}_e \varepsilon_e, \quad (2.10)$$

where $\mathbf{D}_e$ is the elasticity matrix.

Force–displacement relational expression

In an element, the virtual work principle is expressed by

$$\delta U_e = \delta W_e, \quad (2.11)$$

where δU_e is virtual energy, and δW_e is virtual work.

Virtual energy is calculated by

$$\delta U_e = \int_{V_e} \delta \varepsilon_e \sigma_e dV_e, \quad (2.12)$$

where V_e is the volume of the tetrahedral element. As $\delta \varepsilon_e = \mathbf{B}_e \delta \mathbf{u}_e$ according to Eq. (2.7) and Eq. (2.10), Eq. (2.12) is rewritten as

$$\begin{aligned} \delta U_e &= \int_{V_e} \mathbf{B}_e \{ \delta \mathbf{u}_e \} \mathbf{D}_e \mathbf{B}_e \mathbf{u}_e dV_e \\ &= \{ \delta \mathbf{u}_e \}^T \int_{V_e} \mathbf{B}_e^T \mathbf{D}_e \mathbf{B}_e \mathbf{u}_e dV_e. \end{aligned} \quad (2.13)$$

Virtual work is calculated by

$$\delta W_e = \{ \delta \mathbf{u}_e \}^T \mathbf{f}_e, \quad (2.14)$$

where $\mathbf{f}_e$ is the external force vector of nodes of an element.

According to Eq. (2.11), Eq. (2.13) is equal to Eq. (2.14). Therefore,

$$\mathbf{f}_e = \mathbf{K}_e \mathbf{u}_e, \quad (2.15)$$

where $\mathbf{K}_e$ is the elemental stiffness matrix and is calculated by

$$\mathbf{K}_e = \int_{V_e} \mathbf{B}_e^T \mathbf{D}_e \mathbf{B}_e dV_e. \quad (2.16)$$

As the elements are primary tetrahedral elements, Eq. (2.16) is rewritten to

$$\mathbf{K}_e = \mathbf{B}_e^T \mathbf{D}_e \mathbf{B}_e V_e. \quad (2.17)$$

V_e only depends on the coordinates of four nodes. In Fig. 2.5, it can be calculated using the following equation.

$$V_e = \frac{1}{6} \begin{vmatrix} 1 & 1 & 1 & 1 \\ x_1 & x_2 & x_3 & x_4 \\ y_1 & y_2 & y_3 & y_4 \\ z_1 & z_2 & z_3 & z_4 \end{vmatrix} \quad (2.18)$$

The element level of the virtual work principle described so far extended to the whole object. Therefore, it is possible to derive the total stiffness matrix by comprising multiple elements. As a result, the stiffness equation is

$$\mathbf{f} = \mathbf{K}\mathbf{u}, \quad (2.19)$$

where $\mathbf{f}$ and $\mathbf{u}$ are respectively the external force and displacement vectors of total nodes. The total stiffness matrix is calculated by

$$\mathbf{K} = \sum_{e=1}^{\text{all}} \mathbf{A}_e \mathbf{K}_e, \quad (2.20)$$

where $\sum_{e=1}^{\text{all}} \mathbf{A}_e$ is a finite element assembly operator.

2.3.2 Dynamic motion equation of FEM

Dynamic motion equation

As the actual brain tissue is a viscoelastic body, the effect of speed and acceleration of surgical instruments is significant. Therefore, the dynamic behavior of the brain tissue should be expressed. Equation (2.19) represents a static motion equation. To obtain the dynamic law of motion, an inertia term and a damping term should be added to Eq. (2.19). Therefore, the following dynamic motion equation is obtained.

$$\mathbf{f} = \mathbf{M}\ddot{\mathbf{u}} + \mathbf{C}\dot{\mathbf{u}} + \mathbf{K}\mathbf{u}, \quad (2.21)$$

where $\mathbf{f}$ is the external force vector. $\mathbf{u}$, $\dot{\mathbf{u}}$, and $\ddot{\mathbf{u}}$ are the displacement, velocity, and acceleration vectors of nodes, respectively. $\mathbf{K}$, $\mathbf{C}$, and $\mathbf{M}$ are the stiffness, damping, and mass matrices, respectively.

In this study, $\mathbf{C}$ is assumed to be the following Rayleigh damping,

$$\mathbf{C} = a\mathbf{M} + b\mathbf{K}, \quad (2.22)$$

where a and b are Rayleigh damping coefficients.

For the dynamic analysis of the continuum using the FEM, a mass matrix can be used with two methods: the lumped mass (LM) matrix method and consistent mass (CM) matrix method. The LM method entails significant computational advantages

for calculations involving $\mathbf{M}^{-1}$ because it is a diagonal matrix. In contrast, the CM method is preferred for accuracy. However, if the size of elements is not small enough, the accuracy of the CM method is not necessarily better than that of the LM method. Therefore, an LM matrix is adopted. The LM matrix is constructed by distributing the mass to nodes. Therefore, the mass matrix is assumed as

$$\mathbf{M} = \begin{pmatrix} \mathbf{m}_1 & & & \\ & \ddots & & \mathbf{0} \\ & & \mathbf{m}_i & \\ & \mathbf{0} & & \ddots \\ & & & & \mathbf{m}_n \end{pmatrix}, \quad (2.23)$$

where

$$\mathbf{m}_i = \begin{pmatrix} m_i & 0 & 0 \\ 0 & m_i & 0 \\ 0 & 0 & m_i \end{pmatrix}, \quad (2.24)$$

$$m_i = \sum_{j=1}^{N_{ij}} \rho V_j / 4, \quad (2.25)$$

$\mathbf{m}_i$ is the part of the lumped mass of node i , ρ is the density, V_j is the volume of element j , and N_{ij} is the number of elements that share node i . Based on Eq. (2.23), the mass matrix $\mathbf{M}$ is a diagonal matrix.

Numerical Integration

Numerical integration method is generally used to analyze time history of the structure, and is mainly divided into implicit and explicit methods. Computational cost of the explicit method is lower than that of the implicit method. However, to obtain a stable solution for the explicit method, the sampling time must be lower than a certain value. Therefore, the explicit method requires enormous computational steps and is not suitable for a real-time simulator. In this study, the implicit method is adopted.

In implicit methods, a Newmark- β method [63] is commonly used, in which the relationships of displacement $\mathbf{u}$, velocity $\dot{\mathbf{u}}$, and acceleration $\ddot{\mathbf{u}}$ at $n+1$ and n steps are

assumed as follows:

$$\dot{\mathbf{u}}^{(n+1)} = \dot{\mathbf{u}}^{(n)} + \Delta t \left(\gamma \ddot{\mathbf{u}}^{(n+1)} + (1 - \gamma) \ddot{\mathbf{u}}^{(n)} \right), \quad (2.26)$$

$$\mathbf{u}^{(n+1)} = \mathbf{u}^{(n)} + \Delta t \dot{\mathbf{u}}^{(n)} + \Delta t^2 \left(\left(\frac{1}{2} - \beta \right) \ddot{\mathbf{u}}^{(n)} + \beta \ddot{\mathbf{u}}^{(n+1)} \right), \quad (2.27)$$

where β and γ are factors of the Newmark method, and $\Delta t = t^{(n+1)} - t^{(n)}$.

Generally, $\gamma = 1/2$. Moreover, when $\beta \geq 1/4$, the solution is stable without divergence. Therefore, in this study, $\beta = 1/4$ and $\gamma = 1/2$ are adopted. In this case, acceleration is assumed as an average constant value from steps n to $n + 1$. This is commonly called as the trapezoidal rule. Equations (2.26) and (2.27) can be rewritten as

$$\dot{\mathbf{u}}^{(n+1)} = \dot{\mathbf{u}}^{(n)} + \frac{\Delta t}{2} \left(\ddot{\mathbf{u}}^{(n+1)} + \ddot{\mathbf{u}}^{(n)} \right), \quad (2.28)$$

$$\mathbf{u}^{(n+1)} = \mathbf{u}^{(n)} + \Delta t \dot{\mathbf{u}}^{(n)} + \frac{\Delta t^2}{4} \left(\ddot{\mathbf{u}}^{(n)} + \ddot{\mathbf{u}}^{(n+1)} \right), \quad (2.29)$$

Equations (2.28) and (2.29) are motified to express $\dot{\mathbf{u}}^{(n+1)}$ and $\ddot{\mathbf{u}}^{(n+1)}$ by using $\mathbf{u}^{(n+1)}$, $\mathbf{u}^{(n)}$, $\dot{\mathbf{u}}^{(n)}$, and $\ddot{\mathbf{u}}^{(n)}$, as follows

$$\dot{\mathbf{u}}^{(n+1)} = a_0 \left(\mathbf{u}^{(n+1)} - \mathbf{u}^{(n)} \right) - a_1 \dot{\mathbf{u}}^{(n)} - a_2 \ddot{\mathbf{u}}^{(n)}, \quad (2.30)$$

$$\ddot{\mathbf{u}}^{(n+1)} = a_3 \left(\mathbf{u}^{(n+1)} - \mathbf{u}^{(n)} \right) - a_4 \dot{\mathbf{u}}^{(n)} - a_5 \ddot{\mathbf{u}}^{(n)}, \quad (2.31)$$

where

$$a_0 = \frac{\gamma}{\beta \Delta t}, \quad a_1 = \frac{1}{2\beta} - 1, \quad a_2 = \left(\frac{\gamma}{2\beta} - 1 \right) \Delta t, \\ a_3 = \frac{1}{\beta \Delta t^2}, \quad a_4 = \frac{1}{\beta \Delta t}, \quad a_5 = \frac{\gamma}{2\beta} - 1.$$

2.3.3 Calculation of deformation and force in the dynamic FEM

The nodes in a FE model can be divided into two states as contact and free nodes. Contact nodes are the nodes in contact with other objects and are provided forced displacement or boundary conditions. Free nodes are the nodes that have no constraint and move freely.

In this study, knowns and unknowns are different in contact and free nodes, as illustrated in Fig. 2.6. Contact nodes (“○” in Fig. 2.6) that contact with surgical instruments

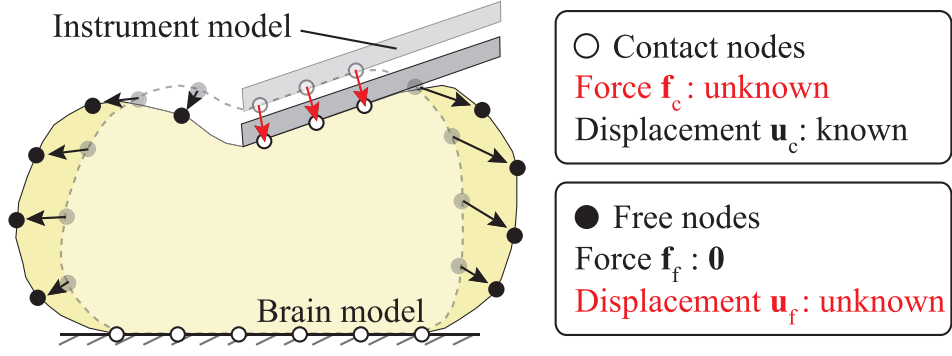


Figure 2.6: Definition of known and unknown amount by the nodal condition.

or stage are assumed to move together with the surgical instruments and stage. For contact nodes, forced displacement is an input so that the displacement is known, while force is unknown. In contrast, for free nodes (“●” in Fig. 2.6), force is known to be $\mathbf{0}$, while displacement is unknown. Therefore, force and displacement for contact nodes and free nodes are calculated separately.

Based on Eq. (2.21), the dynamic motion equation at step $n + 1$ is written as

$$\mathbf{f}^{(n+1)} = \mathbf{M}\ddot{\mathbf{u}}^{(n+1)} + \mathbf{C}\dot{\mathbf{u}}^{(n+1)} + \mathbf{K}\mathbf{u}^{(n+1)}. \quad (2.32)$$

Equation (2.32) is separated into contact nodes and free nodes is as follows:

$$\begin{bmatrix} \mathbf{f}_c^{(n+1)} \\ \mathbf{f}_f^{(n+1)} \end{bmatrix} = \begin{bmatrix} \mathbf{K}_{cc} & \mathbf{K}_{cf} \\ \mathbf{K}_{fc} & \mathbf{K}_{ff} \end{bmatrix} \begin{bmatrix} \mathbf{u}_c^{(n+1)} \\ \mathbf{u}_f^{(n+1)} \end{bmatrix} + \begin{bmatrix} \mathbf{C}_{cc} & \mathbf{C}_{cf} \\ \mathbf{C}_{fc} & \mathbf{C}_{ff} \end{bmatrix} \begin{bmatrix} \dot{\mathbf{u}}_c^{(n+1)} \\ \dot{\mathbf{u}}_f^{(n+1)} \end{bmatrix} + \begin{bmatrix} \mathbf{M}_c & \mathbf{0} \\ \mathbf{0} & \mathbf{M}_f \end{bmatrix} \begin{bmatrix} \ddot{\mathbf{u}}_c^{(n+1)} \\ \ddot{\mathbf{u}}_f^{(n+1)} \end{bmatrix}, \quad (2.33)$$

where subscript “c” expresses contact–node–related terms, and subscript “f” expresses free–node–related terms.

The detail of calculation of unknowns ($\mathbf{u}_f^{(n+1)}$ and $\mathbf{f}_c^{(n+1)}$) is shown as follows.

1. Displacement of contact nodes

Forced displacement is provided to contact nodes as follows:

$$\mathbf{u}_c^{(n+1)} \quad (2.34)$$

and is stored.

2. Velocity and acceleration of contact nodes

As $\mathbf{u}_c^{(n+1)}$ is known, according to Eq. (2.30) and (2.31), $\dot{\mathbf{u}}_c^{(n+1)}$ and $\ddot{\mathbf{u}}_c^{(n+1)}$ can be

calculated as

$$\dot{\mathbf{u}}_c^{(n+1)} = a_0 \left(\mathbf{u}_c^{(n+1)} - \mathbf{u}_c^{(n)} \right) - a_1 \dot{\mathbf{u}}_c^{(n)} - a_2 \ddot{\mathbf{u}}_c^{(n)}, \quad (2.35)$$

$$\ddot{\mathbf{u}}_c^{(n+1)} = a_3 \left(\mathbf{u}_c^{(n+1)} - \mathbf{u}_c^{(n)} \right) - a_4 \dot{\mathbf{u}}_c^{(n)} - a_5 \ddot{\mathbf{u}}_c^{(n)}. \quad (2.36)$$

3. Displacement of free nodes

As $\mathbf{f}_f^{(n+1)} = \mathbf{0}$, Eq. (2.33) can be rewritten as

$$\begin{aligned} \mathbf{u}_f^{(n+1)} = & \left[\mathbf{K}_{ff} + a_0 \mathbf{C}_{ff} + a_3 \mathbf{M}_f \right]^{-1} \left[\mathbf{C}_{ff} \left(a_0 \mathbf{u}_f^{(n)} + a_1 \dot{\mathbf{u}}_f^{(n)} + a_2 \ddot{\mathbf{u}}_f^{(n)} \right) \right. \\ & \left. + \mathbf{M}_f \left(a_3 \mathbf{u}_f^{(n)} + a_4 \dot{\mathbf{u}}_f^{(n)} + a_5 \ddot{\mathbf{u}}_f^{(n)} \right) - \mathbf{K}_{fc} \cdot \mathbf{u}_c^{(n+1)} - \mathbf{C}_{fc} \cdot \dot{\mathbf{u}}_c^{(n+1)} \right] \end{aligned} \quad (2.37)$$

$\mathbf{u}_c^{(n+1)}$ and $\dot{\mathbf{u}}_c^{(n+1)}$ are knowns; therefore, $\mathbf{u}_f^{(n+1)}$ can be calculated.

4. Velocity and acceleration of free nodes

By using Eq. (2.30) and (2.31), $\dot{\mathbf{u}}_f^{(n+1)}$ and $\ddot{\mathbf{u}}_f^{(n+1)}$ can be calculated as

$$\dot{\mathbf{u}}_f^{(n+1)} = a_0 \left(\mathbf{u}_f^{(n+1)} - \mathbf{u}_f^{(n)} \right) - a_1 \dot{\mathbf{u}}_f^{(n)} - a_2 \ddot{\mathbf{u}}_f^{(n)}, \quad (2.38)$$

$$\ddot{\mathbf{u}}_f^{(n+1)} = a_3 \left(\mathbf{u}_f^{(n+1)} - \mathbf{u}_f^{(n)} \right) - a_4 \dot{\mathbf{u}}_f^{(n)} - a_5 \ddot{\mathbf{u}}_f^{(n)}. \quad (2.39)$$

5. External force of contact nodes

By substituting Eqs. (2.34), (2.35), (2.36), (2.37) and (2.38) into Eq. (2.33), $\mathbf{f}_c^{(n+1)}$ can be calculated as follows.

$$\mathbf{f}_c^{(n+1)} = \mathbf{K}_{cc} \cdot \mathbf{u}_c^{(n+1)} + \mathbf{C}_{cc} \cdot \dot{\mathbf{u}}_c^{(n+1)} + \mathbf{M}_c \cdot \ddot{\mathbf{u}}_c^{(n+1)} + \mathbf{K}_{cf} \cdot \mathbf{u}_f^{(n+1)} + \mathbf{C}_{cf} \cdot \dot{\mathbf{u}}_f^{(n+1)} \quad (2.40)$$

As described earlier, the deformation and external force of all the nodes are calculated. Unknowns $\mathbf{u}_f^{(n+1)}$ and $\mathbf{f}_c^{(n+1)}$ are calculated using Eq. (2.37) and Eq. (2.40), respectively. The computing procedure is shown in Fig. 2.7.

The reaction force $\mathbf{f}_{re}$ of an area is calculated by the sum of force of nodes in the area:

$$\mathbf{f}_{re} = \sum_{i=1}^{N_c} \mathbf{f}_{ci}, \quad (2.41)$$

where N_c is the number of contact nodes in the area and $\mathbf{f}_{ci}$ is the external force of contact node i .

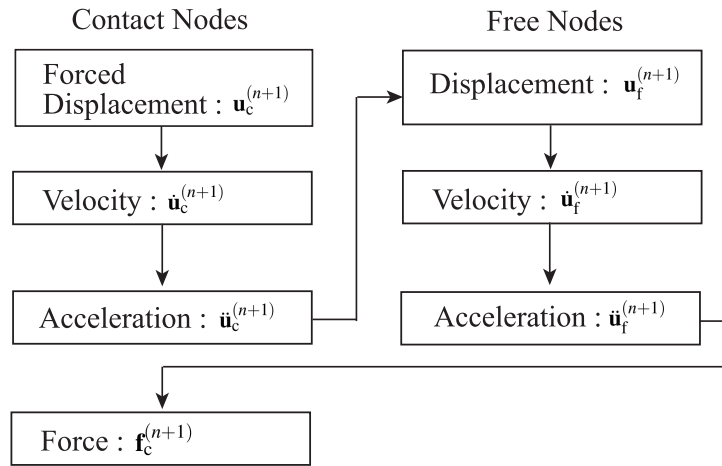


Figure 2.7: Process of calculation of the dynamic motion equation.

2.4 Measurement of Mechanical Properties [64]

The parameters to identify for the model are density ρ (in Eq. (2.25)), Young's modulus E (in Eq. (2.9)), Poisson's ratio ν (in Eq. (2.9)), and Rayleigh damping coefficients a and b (in Eq. (2.22)).

Rayleigh damping is used to stabilize the calculation and it does not reproduce the actual phenomenon. Therefore, the Rayleigh damping coefficients do not have a physical meaning. There are many studies on the measurement of Young's modulus [53, 54]. However, there are large disparities among their identified values because the studies used different testing protocols at different situations [52, 51]. Moreover, in most of the earlier studies, elastic characterization included the influence of strain rate [54]. Therefore, Rayleigh damping coefficients and Young's modulus of brain tissue in state condition should be measured.

Rashid *et al.* presented the values of Poisson's ratio and density of the brain tissue [54]. However, they did not cite scientific evidence about the value. In this study, we measured Poisson's ratio and density of the brain tissue.

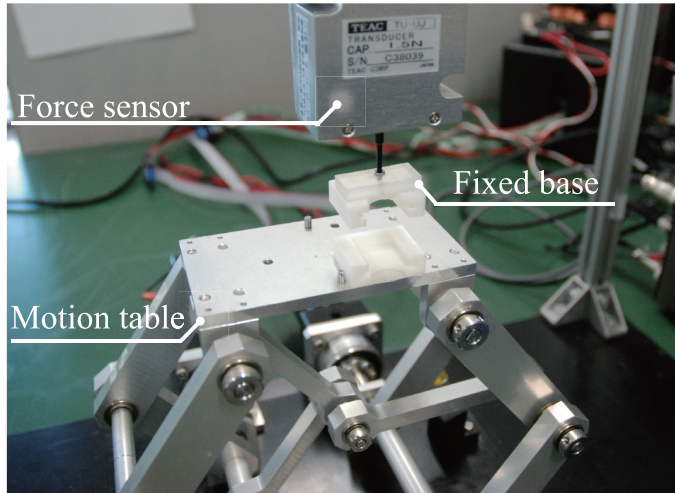


Figure 2.8: Equipment of measurement system

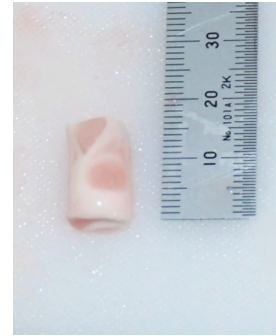


Figure 2.9: Example of specimens

2.4.1 Specimen and equipment

Specimen Preparation

Porcine brains were used to conduct experiments. The swines were roughly six months old and were destroyed on the experiment day. Brains were taken out from heads just before the experiment and tested within 12 h postmortem.

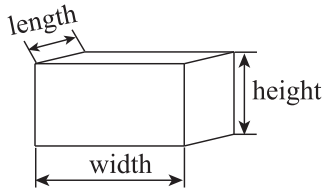
Brains and specimens were preserved in physiological saline solution, which was applied to the specimens frequently during cutting and before testing to prevent dehydration. The laboratory temperature was 25°C.

Specimens were cut out from different parts of the brain tissue. Their sizes varied because the specimens were cut by hand. Specimens were mixtures of white and gray matters, as shown in Fig. 2.9; therefore, compositions of each specimen were different. As a result, there were individual differences among specimens.

Equipment

A motion table was used to pull the bottom of the specimens downward (negative direction of z-axis). Both ends of specimens were glued to plastic plates by using instant glue (cyanoacrylate). One plate was set on the top platen attached to the load cell (TEAC

Table 2.1: Sizes of specimens for density measurement.



Specimen No.	Width (mm)	Length (mm)	Height (mm)	Weight (g)
1	15	18	11	3.0
2	16	20	10	3.6
3	16	28	8	3.9
4	16	24	9	3.7
5	22	33	10	7.3
6	20	30	10	6.1
7	17	22	8	3.3

Corp. TU-UJ1.5N), and the other plate was set on a lower platen attached to the motion table. Before starting each test, the motion table was carefully moved to determine initial position so that the specimen was of its natural length at beginning. We waited one minute after setting to ensure adhesion between the specimen and platens. Fig. 2.8 shows the equipment system of the measurement.

2.4.2 Measurement of density

Seven cuboid specimens were used to measure the density. First, the length, width, and height of specimens were measured to calculate their volumes. Then, the weight of specimens was measured using an electronic balance. The measurement accuracy of electronic balance is 0.1 (g). Table 2.1 lists the measurement results.

The weight and volume are plotted in Fig. 2.10. The density of brain tissue is identified by linear approximation, with the following result.

$$\text{weight} = 1036.4 \cdot \text{volume},$$

$$R^2 = 0.9849.$$

Therefore, density of brain tissue is 1036.4 (kg/m³).

The density of the brain is reported to be close to that of water [65]. Rashid *et al.* set density of the brain as 1040 (kg/m³) [54], and did not provide a scientific basis. In this study, the measurement result is 1036.4 (kg/m³), which is slightly higher than that

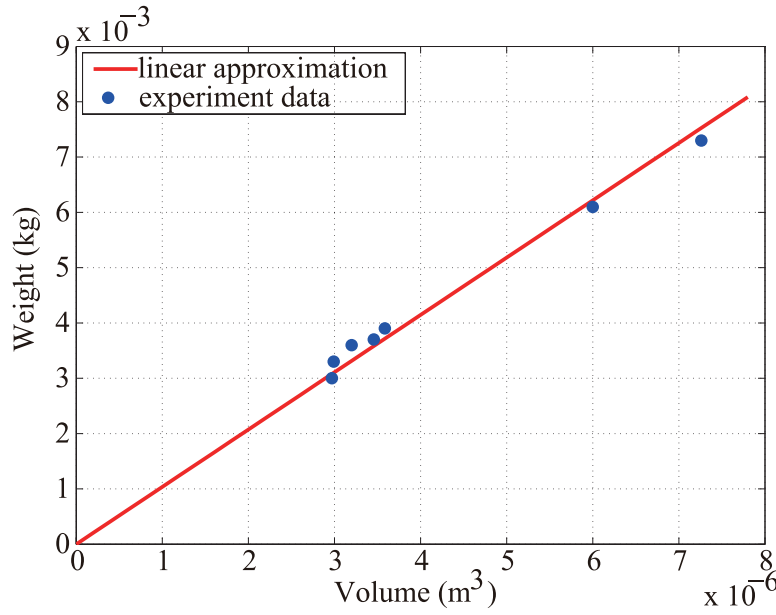


Figure 2.10: Measurement results of density and linear approximation.

of water. Although there are some individual differences, the value is reliable.

2.4.3 Measurement of Poisson's ratio

In the present study, the specimens were cut into cuboids to measure Poisson's ratio. Poisson's ratio is calculated as:

$$\nu = -\frac{\varepsilon_x}{\varepsilon_z}, \quad (2.42)$$

where ε_x and ε_z are strains along the x -axis and z -axis.

Tensile tests were conducted to measure the Poisson's ratio. Four specimens were used in the tensile tests. Four markers were affixed to a specimen. The velocity of the motion table was 0.1 (mm/s). Simultaneously, the motion was captured by a high-speed camera to record the movement of the markers, as shown in Fig. 2.11. The frame rate of the camera is 100 (fps).

The z -axis strain is calculated by the displacement of the motion table Δh and the initial height of specimens h as

$$\varepsilon_z = \frac{\Delta h}{h}.$$

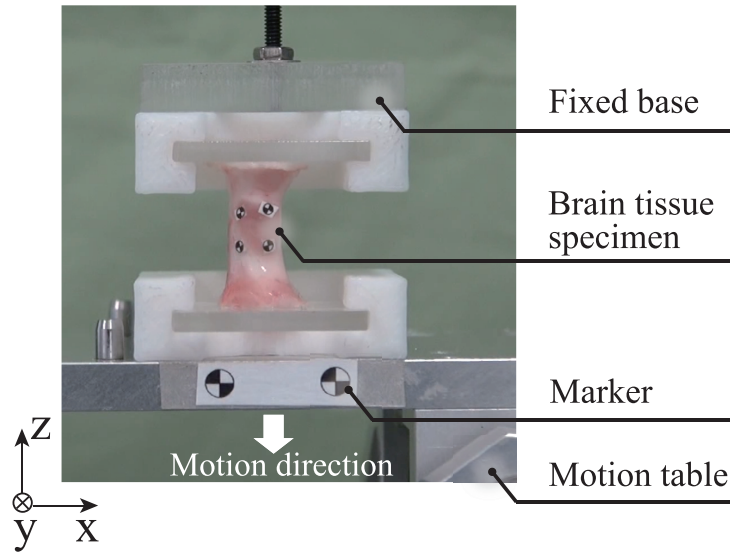


Figure 2.11: Equipment for measuring Poisson's ratio.

The x -axis strain is calculated by the x -axis distance between the markers. Two x -axis strains are measured using four markers, as shown in Fig. 2.12. The initial widths are w_1 and w_2 . The average of the two strains is used. Therefore, ϵ_x is calculated as

$$\epsilon_x = \frac{1}{2} \left(\frac{\Delta w_1}{w_1} + \frac{\Delta w_2}{w_2} \right).$$

An example of $-x$ -axis strain measurement by using the high-speed camera is plotted in Fig. 2.13. $-x$ -axis and z -axis strains are plotted in Fig. 2.14. Moreover, the Poisson's ratio of the brain tissue is identified through linear approximation, with the following result.

$$\begin{aligned} -\epsilon_x &= 0.4925 \cdot \epsilon_z, \\ R^2 &= 0.9957. \end{aligned}$$

Therefore, Poisson's ratio of the brain tissue is 0.4925.

The brain is an almost incompressible gel-like material; therefore, Poisson's ratio is close to 0.5. Rashid *et al.* set Poisson's ratio as 0.4999 [54]. In the present study, the Poisson's ratio is experimentally identified to be 0.4925, which is close to the value presented in [54]. Therefore, the value is reliable.

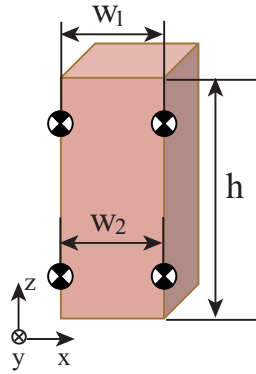


Figure 2.12: Illustrator of specimen for Poisson's ratio measurment.

Table 2.2: Measurment results of Poisson's ratio.

Specimen	ϵ_z	$-\epsilon_x$
1	0.0766	0.0376
2	0.2857	0.1427
3	0.0750	0.0367
4	0.0976	0.0425
5	0.2229	0.1540

2.5 Summary

The details of FE model of brain tissue were presented in this chapter. The geometric model of brain tissue was created. In addition, the deformation and force of the object model were calculated using a dynamic FEM. Moreover, the density and Poisson's ratio which are necessary for the numerical model were measured.

Visual and force information were generated from the numerical model. Thus, information can be provided to operators. Therefore, the arrangement of the neurosurgery simulation was completed.

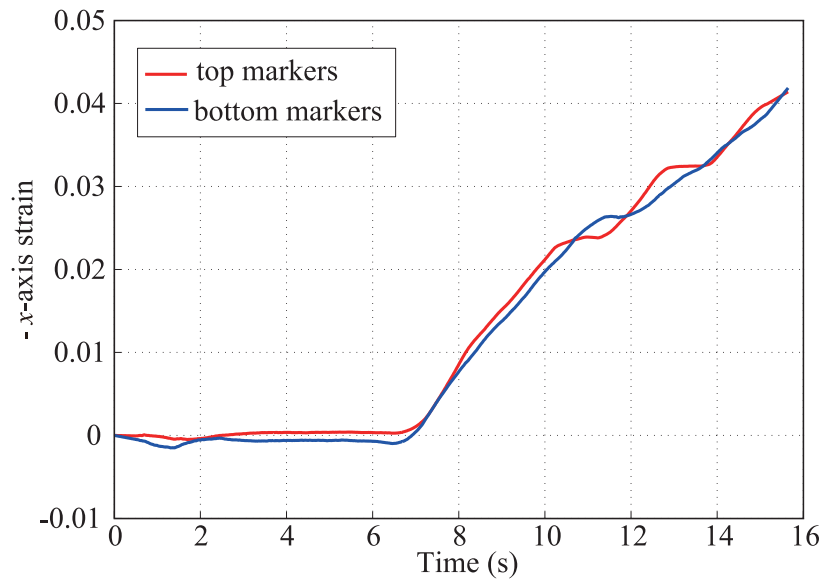


Figure 2.13: Example of x -axis strain measurement results.

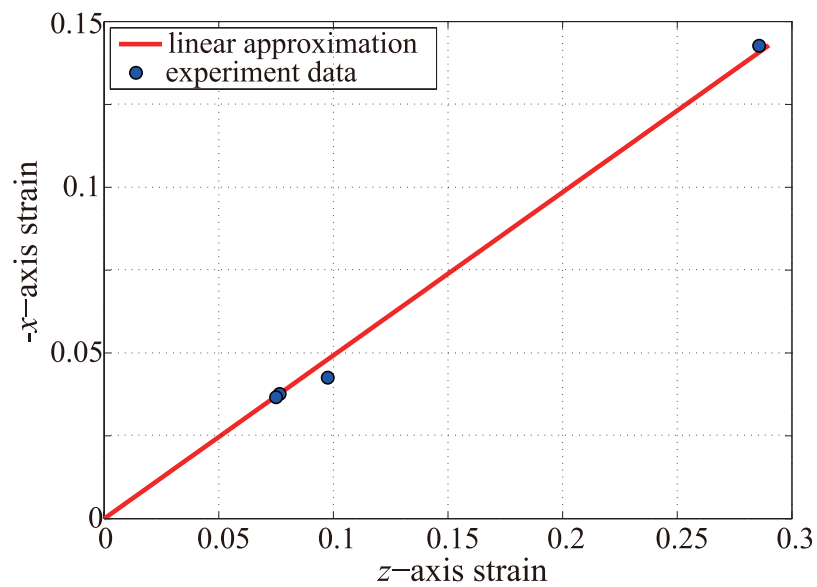


Figure 2.14: Measurement results of Poisson's ratio and linear approximation.

Chapter 3. Damage Fracture Model

3.1 Introduction

In this chapter, the details of tensile tests and blunt dissection simulation are presented. The main goal of this chapter is to develop damage fracture models of brain parenchyma for haptic brain surgery simulations. Dissection and removal of lesion areas are fundamental operations in brain surgeries. Therefore, damage and fracture models are needed to simulate dissection and removal operations. Generally, brain tissues show strong ductility; however, conventional fracture or damage models cannot satisfactorily reproduce ductile fractures.

In this chapter, to ensure damage and fracture mechanics of brain tissues, tensile tests with two strain rates were conducted using porcine brain parenchyma, with tensile velocities 0.1 (mm/s) and 1.0 (mm/s).

Moreover, two simple damage and fracture models of brain parenchyma are proposed for real-time haptic surgery simulations. In model A, it is assumed that micro-damage begins because of the von Mises yield criterion, and its growth rate is in proportion to its volume strain. In model B, it is assumed that micro-damage begins when von Mises stress exceeds a certain threshold in an element, and the micro-damage grows with increased von Mises stress. The stiffness decreases with the increasing micro-damage. When the integrity of an element reduces below a certain threshold, the element is removed to express the occurrence of a fracture. These steps were formulated algorithmically. Although the proposed model does not require iterative calculation, it can reproduce ductile fracture while maintaining sufficient accuracy. The proposed models are suitable for real-time computation. Mechanical properties and proposed damage model parameters were identified for each model by solving an optimization problem with a fitted curve of experimental data at 0.1 (mm/s).

In addition, tensile simulations were performed using the proposed damage and frac-

ture models and identified properties with two tensile velocities (0.1 (mm/s) and 1.0 (mm/s)). The simulation results were compared with the experimental results to verify the proposed damage and fracture models.

This chapter is organized as follows. Section 3.2 presents tensile tests. In Section 3.3, conventional fracture or damage models are used to represent the damage and fracture mechanics of brain tissues and their failure. In Section 3.4 and 3.5, two damage and fracture models are proposed. Moreover, mechanical properties and parameters of the proposed damage models are identified. Section 3.6 describes the simulations.

3.2 Tensile Tests of Porcine Brain Parenchyma

To investigate fracture mechanics of brain tissue, tensile tests were performed. In the tensile tests, porcine brains were used because the mechanical properties of porcine brain parenchyma are similar to those of the human brain [66]. Uniaxial tension and fracture tests were performed to determine the mechanical characteristics of brain tissue.

The mechanical behaviors of brain tissue have been tested in tension, compression, and shear, with various strain rate [30, 31]. However, most of the earlier studies about damage and fracture mechanics of brain tissue target the traumatic brain injury (TBI) with high strain rate [40, 34] (e.c. $1000 - 3000 \text{ (s}^{-1}\text{)}$ in [34]). The mechanical behavior of brain tissue shows strain–rate dependency. Therefore, experimental results and numerical models of TBI are not applicable to the model simulating neurosurgery, which is usually conducted under low strain rate.

3.2.1 Specimen preparation and equipment

Specimen preparation and equipment were the same as described in Section 2.4.1. Thirteen specimens were taken from different areas of the brain cerebrum. First, the specimens were cut from the brain parenchyma by using a cork borer; thus, specimens were cylindrical. A constriction was then formed at the middle of each specimen so that the specimen was fractured at the constriction in the tensile test. As a result of this formation, each specimen was shaped as a hyperboloid, as shown in Fig. 3.1. Individual differences existed among the specimens because of size variation and different

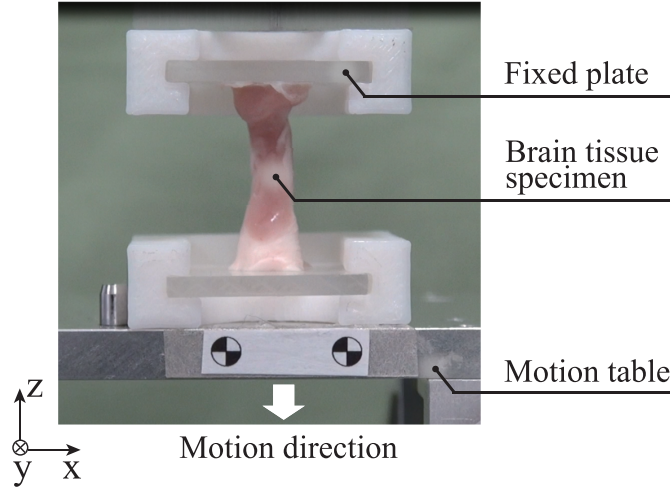


Figure 3.1: Overview of the tensile test system.

composition. Table 3.1 shows the size of specimens.

3.2.2 Results of tensile tests

In the tensile tests, specimens were pulled at two velocities: 0.1 and 1.0 (mm/s). Figure 3.3 shows the progress of how the brain tissue deformed from the initial state to complete separation at a tensile velocity of 0.1 mm/s. Eight successful datasets were obtained: four datasets each at 0.1 and 1.0 (mm/s).

The stress–strain curves at each velocity are plotted in Fig. 3.4. In this study, the stress was calculated using the force and the original end area of the specimens, as shown in Fig. 3.2. In other words, it represents nominal stress. Strain and stress are calculated by

$$\text{strain} = \frac{\Delta L}{L}, \quad (3.1)$$

$$\text{stress} = \frac{F}{S}, \quad (3.2)$$

where L is the initial length of specimens. ΔL is the displacement of motion-table. F was measured by load cell and S is the original end area. As the data obtained from the load cell includes high-frequency noise, a Savitzky–Golay filter (frame size = 100) was used for noise removal. Fitting of a quintic function (a fifth-degree polynomial) was

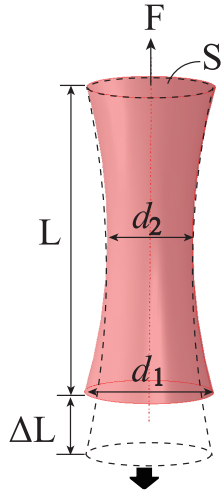


Figure 3.2: Specimen illustration of tensile tests.

Table 3.1: Sizes of specimens of tensile tests.

tensile velocity (mm/s)	Specimen No.	d_1 (mm)	d_2 (mm)	L (mm)
0.1	6	6		15
	8	6	4	16
	10	7		16
	12	6		15
1.0	1	5		14
	4	6	4	13
	7	7		17
	11	6		17
average		6	4	15

performed to the datasets at each velocity. The fitted curve at 0.1 (mm/s) is

$$y = 6683x^5 - 6801x^4 - 6139x^3 + 5108x^2 + 941.9x,$$

$$R^2 = 0.832,$$

and the fitted curve at 1.0 (mm/s) is

$$y = 1470x^5 + 815.3x^4 - 7854x^3 + 4765x^2 + 1029x,$$

$$R^2 = 0.931.$$

The fitted curves are plotted together with the original data in Fig. 3.4.

3.2.3 Differences between brain and other soft biological tissues

The fitted curves for the tensile tests at the fast (1.0 (mm/s)) and slow (0.1 (mm/s)) tension velocities are compared in Fig. 3.5. As shown, both curves are nearly the same when the strain is low. Thereafter, the curve at the high tension velocity reaches a higher ultimate tensile strength and larger strain. This shows that the stress–strain curve for brain tissue depends upon tensile velocity. A similar biomechanical phenomenon has been reported in anterior cruciate ligament [67].

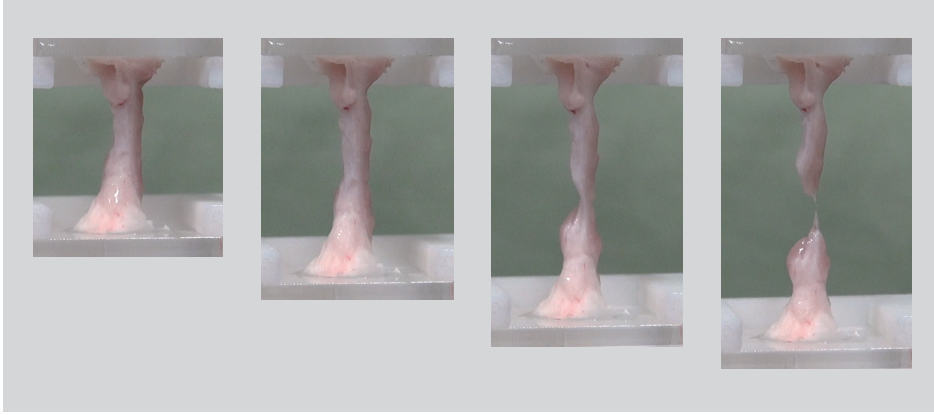
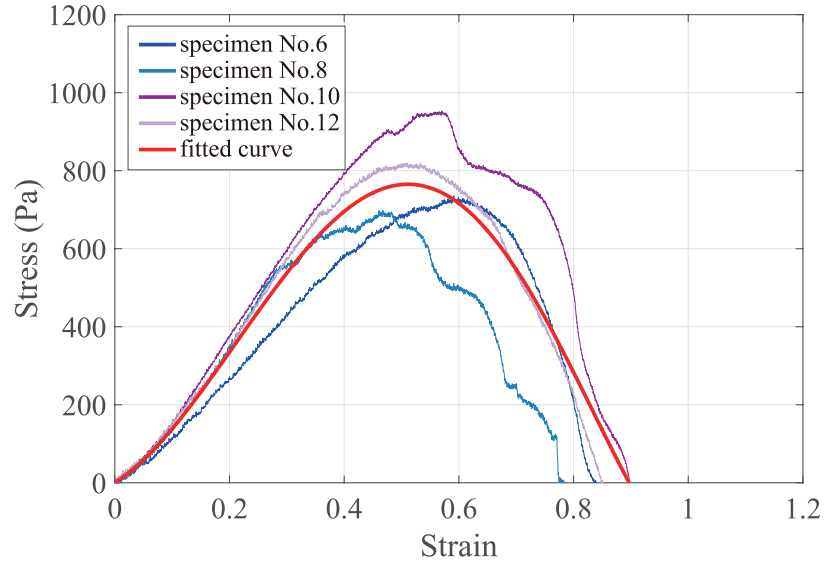


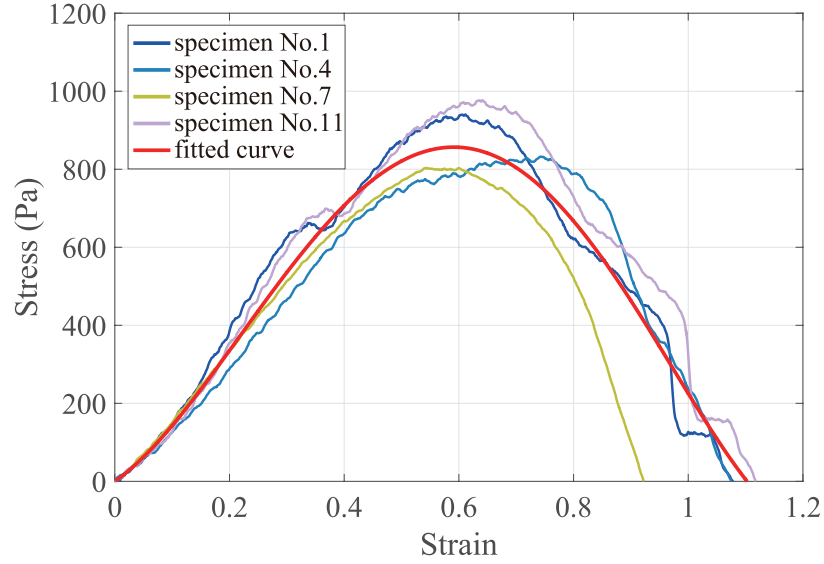
Figure 3.3: Sequences of a tensile test at a velocity of 0.1 mm/s.

Stress–strain curves of other biotissues, such as skin [37], liver [38], and muscle [39] have also been reported. Skin, liver, and muscle showed brittle fracture characteristics as illustrated in Fig. 3.6, whereas brain tissues showed ductile fracture characteristics, as presented in Figs. 3.4 and 3.5.

Although ductile fracture characteristics have been studied in damage mechanics, most researches have been conducted for metals; thus, such researches assumed elastic-plasticity. As shown in Figs. 3.4 and 3.5, explicit elastic-plasticity characteristics are not evident in the stress–strain curves of brain tissue. Therefore, we propose a damage and fracture model for brain parenchyma that can reproduce ductile fracture characteristics.



(a) Tensile velocity: 0.1 mm/s



(b) Tensile velocity: 1.0 mm/s

Figure 3.4: Stress–strain curves of brain tissue in tensile tests.

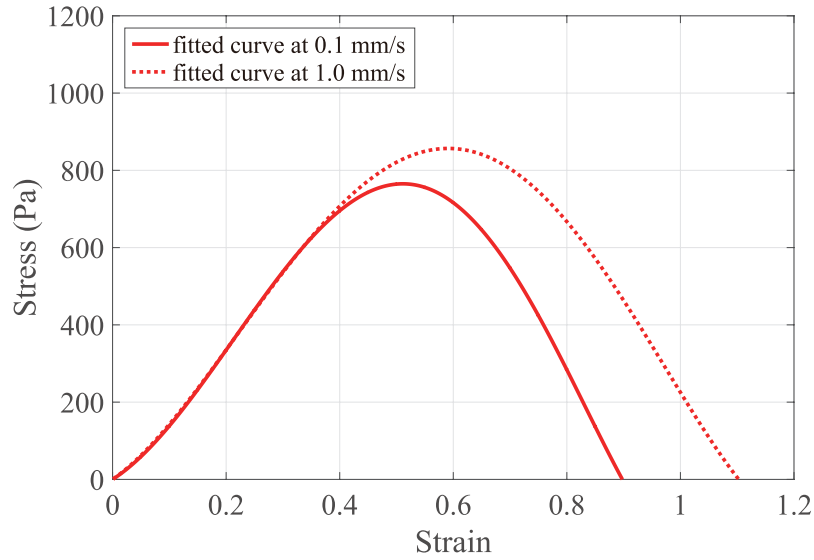


Figure 3.5: Comparison between fitted curves for data at the low tensile velocity (0.1 mm/s) and the high tensile velocity (1.0 mm/s).

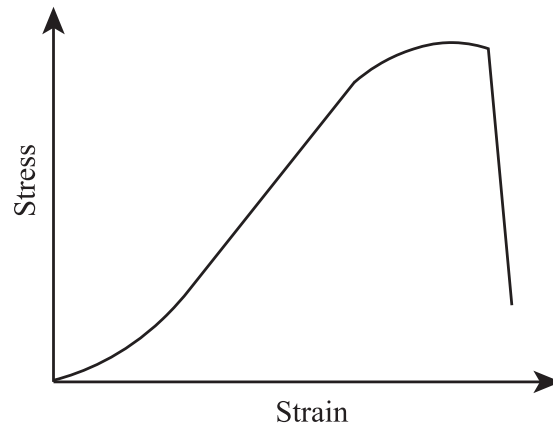


Figure 3.6: Typical stress–strain curve for brittle fracture seen in tensile tests of skin, liver, and muscle tissues.

Table 3.2: Mechanical properties and damage model parameters of conventional fracture or damage models.

Parameters	Fracture model		Damage model	
	RE	XFEM	GT	SL
Density (kg/m^3)	1036			
Poisson's ratio	0.4925			
Sampling time (s)	0.15	0.015	0.015	0.001
Young's modulus (Pa)	2900	3000	3428	2823
Rayleigh damping scalars	$a = 1.0$ $b = 1.0$	-	$a = 1.84$ $b = 1.27$	$a = 1.0$ $b = 1.0$
Threshold of stress (Pa) σ_t	1600	3250 (principal stress)	1000	800
Growth of fracture and damage	-	0.1 (mm) (displacement)	1.2	-

3.3 Previous Studies on Fracture and Damage Numerical Model

Many numerical models have been developed to express damage and fracture of materials. Simulations have been conducted using the conventional fracture and damage models. The models are investigated to check whether they are effective for the brain tissue by comparing the simulation and experiment results. The basic part of simulation is described in Section 2.3.1. The fitted curves in Fig. 3.4 are used for comparison. The size of model in the simulation is used as the average size of specimens, as shown in Table 3.1. Fig. 3.7 illustrates the model. The motion of the simulation is the same as that of tensile tests. The nodes of the upper surface ($z = 15$ (mm)) and bottom surface ($z = 0$ (mm)) are contact nodes. The other nodes are free nodes. The displacement of upper surface nodes is $u_x = u_y = u_z = 0$, and the displacement of bottom surface nodes is $u_x = u_y = 0$, $u_z = -0.1t$ when tensile velocity is 0.1 (mm/s) or $u_x = u_y = 0$, $u_z = -1.0t$ when tensile velocity is 1.0 (mm/s).

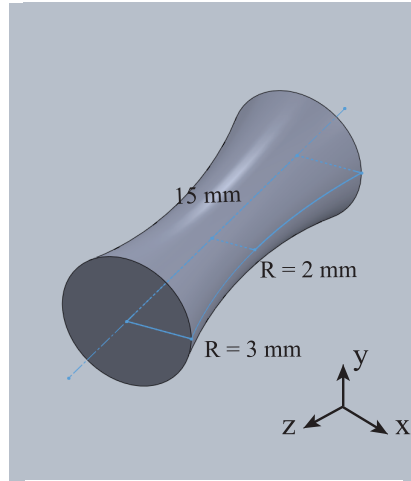


Figure 3.7: Shape and dimensions of the brain tissue model used in the simulation.

3.3.1 Conventional fracture models for biotissues

Removing element method

The simplest method to simulate surgical cuts and dissections is to remove an element [68, 69, 64]. In the removing element (RE) method, the maximum principal stress, shear strain energy, or maximum shear stress of an element are generally used as an index to evaluate whether an element should be removed or not.

A simulation was conducted using an RE method. The condition of removing elements is that the von Mises stress of the elements exceeds a threshold. The stress vector is calculated using Eq. (2.10). The von Mises stress is calculated as follows.

$$\sigma_{Mises} = \sqrt{\frac{1}{2}\{(\sigma_x - \sigma_y)^2 + (\sigma_y - \sigma_z)^2 + (\sigma_x - \sigma_z)^2\} + 3(\tau_{xy}^2 + \tau_{yz}^2 + \tau_{xz}^2)}. \quad (3.3)$$

When the von Mises stress of an element exceeds the threshold, the element stiffness matrix of the element is set as $\mathbf{K}_e = \mathbf{0}$.

Mechanical properties and damage model parameters of simulation with the RE method were identified manually. The identified results are listed in Table 3.2. The tensile velocity is 0.1 (mm/s). The stress–strain curve of the simulation is shown in Fig. 3.8.

In the RE method, several elements tend to reach the fracture stress simultaneously. By comparing the simulation and experimental results (fitted curve in Fig. 3.4 (a)), the RE method is known to reproduce brittle fractures; and therefore it is not an appropriate

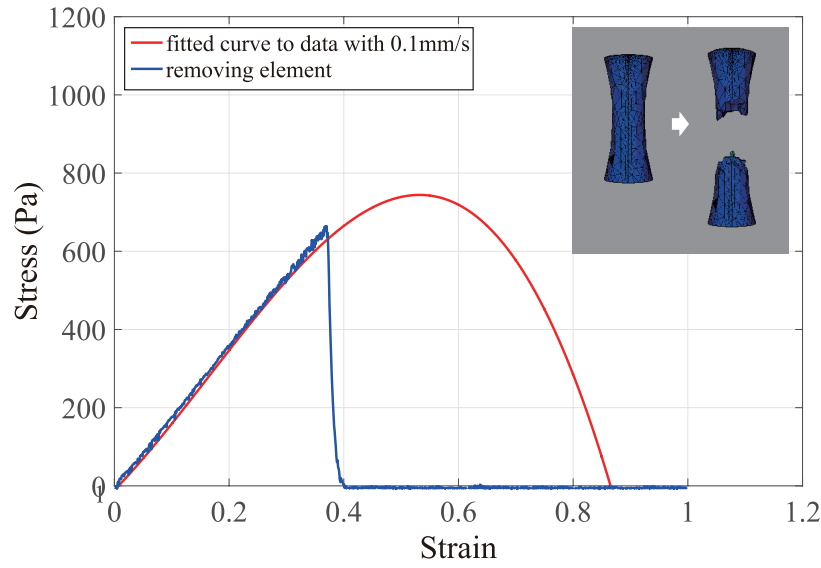


Figure 3.8: Stress–strain curves given by the removing element model.

method to represent fracture mechanics of brain tissue.

Extended finite element method (XFEM)

A considerable number of studies have been conducted on fracture mechanics for metals (e.g., [70]). In the field of computational fracture mechanics, the extended finite element method (XFEM) [71] has been used widely to analyze crack growth. Some researchers have applied XFEM to surgery simulation to simulate surgical cuts and dissections [72, 73]. XFEM makes the modelling possible to model discontinuity easier than conventional FEM.

A tensile test simulation was performed using XFEM. A preset crack was set in the middle of the area $z = 7.5$ (mm). The tensile velocity is 0.1 (mm/s). The obtained stress–strain curve is plotted in Fig. 3.9. ABAQUS/CAE Student Edition 6.14 [74], one of the typical general-purpose FE method analysis software, was used for the XFEM simulation. The maximum principal stress criterion was used to predict damage initiation in the XFEM-enriched region. *Displacement* and *linear* were chosen as the types of damage evolution and the softening method, respectively. Mechanical properties and damage model parameters of simulation were identified manually. The identified results are listed in Table 3.2.

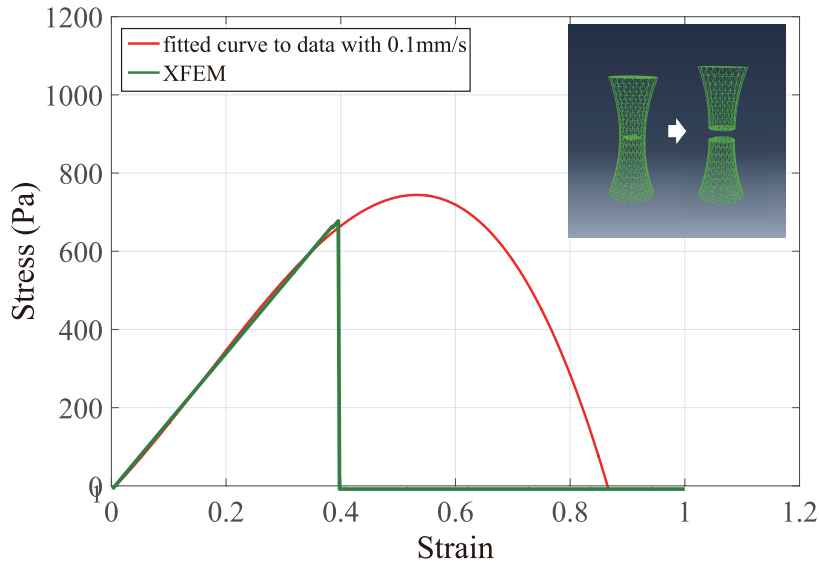


Figure 3.9: Stress—strain curves given by the XFEM analysis using ABAQUS.

In XFEM, the crack grows rapidly after fracture initiation. XFEM effectively simulates brittle fractures, as shown in Fig. 3.9 but not ductile fractures. Therefore, XFEM is not an appropriate method to represent fracture mechanics of brain tissue.

3.3.2 Conventional damage models

Continuum damage mechanics has been studied enthusiastically so far [27]. Generally, a fracture process is considered to consist the nucleation of microscopic cracks and voids, growth and coalescence of microscopic cavities, and resultant macroscopic cracks, as illustrated in Fig. 3.10.

When dealing with microscopic discontinuity effects in continuum mechanics, discontinuous structures should be represented as macroscopic continuous fields by averaging mechanical effects. The minimum volume that satisfies this condition is called a representative volume element (RVE).

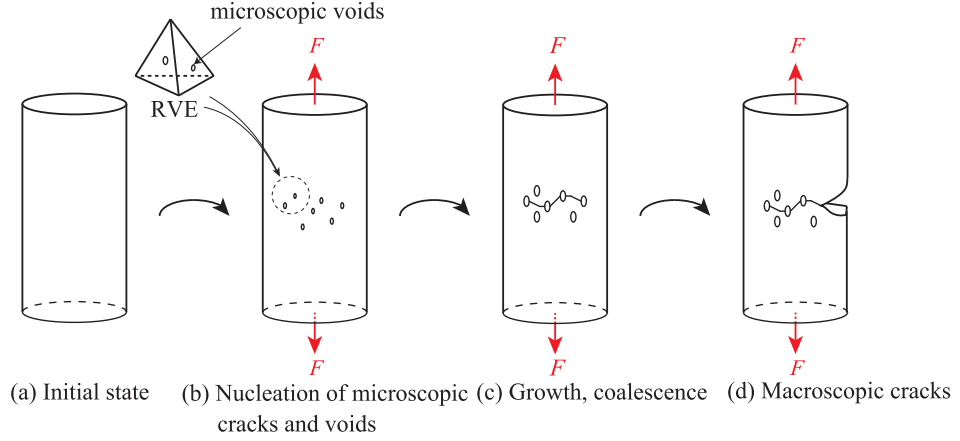


Figure 3.10: Schematic illustration of continuum damage mechanics.

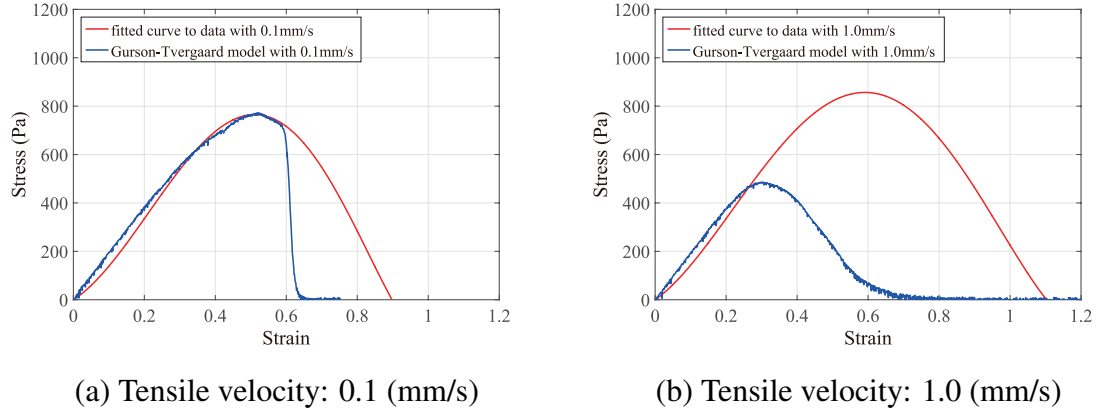


Figure 3.11: Stress–strain curves given by the Gurson–Tvergaard (GT) model.

Gurson–Tvergaard model [75]

Gurson represented the damage state by using the void volume fraction D [29], which is given by

$$D = \frac{V_{\text{void}}}{V_e}, \quad (3.4)$$

where V_e and V_{void} are the volumes of an RVE and the total volume of voids, respectively. Gurson proposed a yield function for a damaged material as follows [29].

$$F(\sigma_{Mises}, \sigma_H, \sigma_y, D) = \left(\frac{\sigma_{Mises}}{\sigma_y} \right)^2 + 2D \cosh \left(\frac{3\sigma_H}{2\sigma_y} \right) - D^2 - 1 = 0, \quad (3.5)$$

where σ_{Mises} , σ_H , and σ_y are von Mises stress, mean stress, and yield stress, respectively. When $D = 0$, Eq. (3.5) corresponds to the von Mises yield criterion [27].

In the Gurson–Tvergaard (GT) model [75], the evolution equation for the void volume fraction is given by

$$\dot{D} = (1 - D)\dot{\epsilon}_{kk}^p + \Lambda\dot{\epsilon}_{eq}^p, \quad (3.6)$$

where $\dot{\epsilon}_{kk}^p$ is the plastic volumetric strain rate and $\dot{\epsilon}_{eq}^p$ is the equivalent plastic strain rate. Λ is a coefficient.

Most continuum damage mechanics models (e.g., in [76, 29, 28]) are based on elastic-plastic theory. Therefore, nonlinear solutions (e.g., using the Newton–Raphson scheme) are needed by invoking iterative calculations. However, such iterative calculation methods are not suitable for haptic surgery simulations because reaction force, deformation, and dissection must be calculated in real time. Furthermore, iterative calculation cannot guarantee the convergence of a solution at a critical time.

Therefore, only the evolution equation for the volume fraction is implemented in a linear dynamic FE model. To evaluate the damage model given by Eq. (3.6), tensile test simulations were performed. The coefficient Λ in Eq. (3.6) is a function of ϵ_{eq}^p in the GT model; however, a constant value is set for Λ to simplify the simulation.

Figure 3.11 (a) shows the stress–strain curve given by tensile test simulation with a tensile velocity of 0.1 (mm/s). Λ and other parameters were provided manually so that the resultant curve reproduced the fitted curve obtained from tensile tests by 0.1 (mm/s), as presented in Section 3.2.

Figure 3.11 (b) shows the stress–strain curve given by tensile simulation with a tensile velocity of 1.0 (mm/s). The same parameters used in the case of the 0.1 (mm/s) velocity were used. Figure 3.11 (b) clearly shows the disadvantage of the damage model given by Eq. (3.6). If the tensile velocity is changed, the damage model does not reproduce the experimental results well.

Simplified Lemaitre Damage Model [77]

The Lemaitre coupled elastic-plasticity-damage model [28] has been extensively used to predict the damage and failure in ductile metals. By removing kinematic hardening from Lemaitre’s original ductile damage theory, a simplified Lemaitre (SL) damage model was obtained [77]. The constitutive equations of the model are listed as follows (see Table 3.3 for variables).

Table 3.3: Characters for simplified Lemaitre damage model

G	shearing modulus	K	bulk modulus
r, s	material constant	$\mathbf{I}$	second-order identity tensor
Φ	yield function	ϵ_v	volumetric strain
ϵ^p	plastic strain tensor	ϵ^e	elastic strain tensor
σ_{y0}	initial yield stress	R	internal variable of isotropic hardening
$\sigma_y(R)$	isotropic hardening stress	Y	damage energy release rate
p	hydrostatic stress	q	equivalent stress
$\dot{\gamma}$	plastic consistency parameter	$\mathbf{s}$	deviatoric stress tensor

(i) Element stiffness matrix

$$\mathbf{K}_e(\omega) = \mathbf{V}_e \mathbf{B}_e^T \mathbf{D}_e(\omega) \mathbf{B}_e \quad (3.7)$$

(ii) Yield function

$$\Phi = \frac{q}{1-D} - \sigma_y \quad (3.8)$$

(iii) Plastic flow and evolution equations for R and D

$$\dot{\epsilon}^p = \dot{\gamma} \sqrt{\frac{3}{2}} \frac{\mathbf{s}}{(1-D)\|\mathbf{s}\|} \quad (3.9)$$

$$\dot{R} = \dot{\gamma} \quad (3.10)$$

$$\dot{D} = \dot{\gamma} \frac{1}{1-D} \left(\frac{-Y}{r} \right)^s \quad (3.11)$$

where

$$Y = -\frac{q^2}{6G(1-D)^2} - \frac{p^2}{2K(1-D)^2} \quad (3.12)$$

(iv) Loading/unloading criterion

$$\Phi \leq 0, \quad \dot{\gamma} \geq 0, \quad \Phi \dot{\gamma} = 0 \quad (3.13)$$

The constitutive equations are highly nonlinear. An algorithm for the numerical integration is presented as follows (see Table 1.1 for variables).

(i) Elastic predictor

$$\boldsymbol{\varepsilon}_{n+1}^{e \text{ trial}} = \boldsymbol{\varepsilon}_n^e + \Delta \boldsymbol{\varepsilon} \quad (3.14)$$

$$R_{n+1}^{\text{trial}} = R_n \quad (3.15)$$

$$\tilde{p}_{n+1} = K \boldsymbol{\varepsilon}_{v n+1}^{e \text{ trial}} \quad (3.16)$$

$$\tilde{\mathbf{s}}_{n+1}^{\text{trial}} = 2G \boldsymbol{\varepsilon}_{d n+1}^{e \text{ trial}} \quad (3.17)$$

$$\tilde{q}_{n+1}^{\text{trial}} = \sqrt{\frac{3}{2}} \|\tilde{\mathbf{s}}_{n+1}^{\text{trial}}\| / (1 - D_n) \quad (3.18)$$

(ii) Check plastic admissibility

$$\begin{aligned} &\text{IF } \tilde{q}_{n+1}^{\text{trial}} - \sigma_y(R_{n+1}^{\text{trial}}) \leq 0 \\ &\text{THEN } (*)_{n+1} = (*)_{n+1}^{\text{trial}} \quad \text{and EXIT} \end{aligned} \quad (3.19)$$

(iii) Return mapping

$$\omega(\Delta\gamma) - \omega_n + \frac{\Delta\gamma}{\omega(\Delta\gamma)} \left(\frac{-Y(\Delta\gamma)}{r} \right)^s = 0 \quad (3.20)$$

where

$$\omega(\Delta\gamma) = \frac{3G\Delta\gamma}{\tilde{q}_{n+1}^{\text{trial}} - \sigma_y(R_n + \Delta\gamma)} \quad (3.21)$$

$$-Y(\Delta\gamma) = \frac{[\sigma_y(R_n + \Delta\gamma)]^2}{6G} + \frac{\tilde{p}_{n+1}^2}{2K} \quad (3.22)$$

The Newton–Raphson method is used to solve this equation for $\Delta\gamma$. The initial guess is

$$\Delta\gamma^{(0)} = \frac{[\tilde{q}_{n+1}^{\text{trial}} - \sigma_y(R_n + \Delta\gamma)]\omega_n}{3G} \quad (3.23)$$

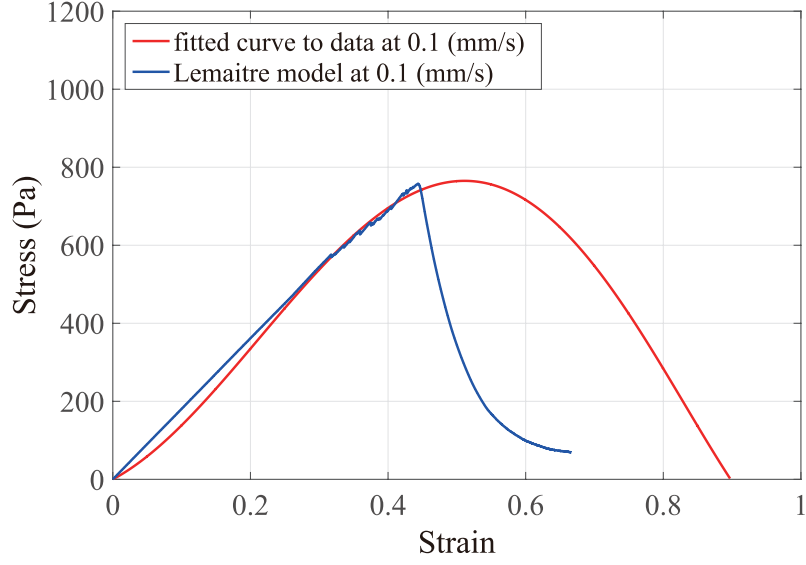


Figure 3.12: A stress–strain curve of a tensile simulation with the simplified Lemaitre damage model.

Then, update

$$R_{n+1} = R_n + \Delta\gamma \quad (3.24)$$

$$p_{n+1} = \omega(\Delta\gamma)\tilde{p}_{n+1} \quad (3.25)$$

$$q_{n+1} = \omega(\Delta\gamma)\sigma_y(R_{n+1}) \quad (3.26)$$

$$\mathbf{s}_{n+1} = \frac{q_{n+1}}{\tilde{q}_{n+1}^{\text{trial}}} \tilde{\mathbf{s}}_{n+1}^{\text{trial}} \quad (3.27)$$

$$\boldsymbol{\sigma}_{n+1} = \mathbf{s}_{n+1} + p_{n+1}\mathbf{I} \quad (3.28)$$

$$\boldsymbol{\epsilon}_{n+1}^e = \frac{1}{2G}\mathbf{s}_{n+1} + \frac{1}{3}\boldsymbol{\epsilon}_{v_{n+1}}^{\text{trial}}\mathbf{I} \quad (3.29)$$

(iv) EXIT

To evaluate the SL model, a tensile simulation was performed. Tensile velocity is 0.1 (mm/s). Mechanical properties and Lemaitre damage model parameters were manually, provided as shown in Table 3.2. The hardening curve is set as $\sigma_y(R) = 800 + 1800(1 - \exp(-0.4R))$, and material properties are $s = 1$ and $r = 0.35$.

Figure 3.12 shows the stress–strain curve of the tensile simulation using the SL model (blue solid line), and it is compared with the fitted curve of tensile tests at 0.1 (mm/s) (red solid line, which is the same as the solid line in Fig. 3.5).

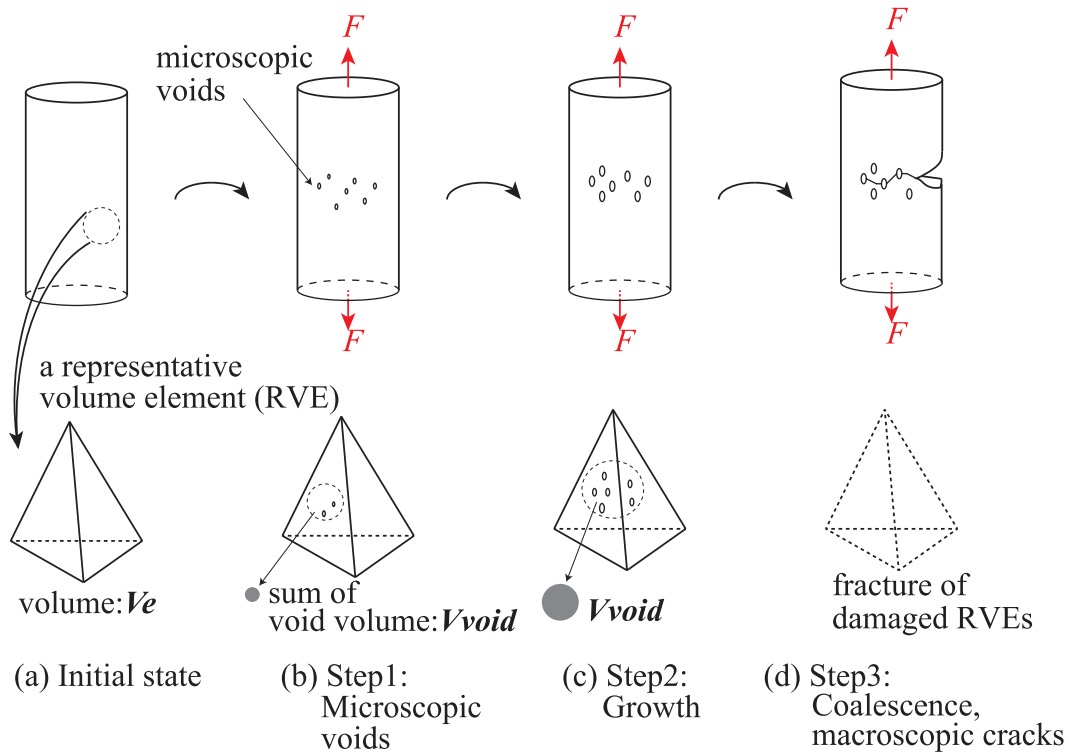


Figure 3.13: Schematic illustration of mechanics of damage model.

As shown in Fig. 3.12, damage occurs in the plastic region, and develops with increasing plastic strain. When the plastic strain reaches a certain value, damage proceeds at a rapid rate and leads to fracture. The damage growth is strongly dependent on the stress triaxiality ratio (p/q) in ductile metals. The phenomenon that damage growth remarkably increases as stress triaxiality ratio increases is captured by the Lemaitre damage model [78]. As a result, although the simulation result in elastic region (the region where the stress and strain are almost linear related) is similar to the experimental results, it is not able to reproduce the ductility of the experimental results by using the SL model, as shown in Fig. 3.12. Therefore, a new damage fracture model should be developed.

3.4 Damage Fracture Model A [79]

This section presents the proposed damage fracture model A based on CDM. Moreover, the parameters required for damage fracture model A are identified.

3.4.1 Damage fracture model A

The concept of the proposed damage model is shown in Fig. 3.13. The damage and fracture process is expressed through three steps: (1) micro-damage initiation, (2) micro-damage growth, and (3) coalescence of micro-damages.

In this model, the integrity ω of the RVE is defined as

$$\begin{cases} \omega = 1 & (\text{corresponds to the undamaged state}), \\ \omega_t < \omega < 1 & (\text{characterizes the damaged state}), \\ \omega \leq \omega_t & (\text{corresponds to a completely damaged state}), \end{cases} \quad (3.30)$$

where ω_t is the threshold for fracture. The relationship of damage variable D and integrity ω is

$$\omega \equiv 1 - D. \quad (3.31)$$

The three steps of fracture process in an element (RVE) are summarized as follows:

Step1: Micro-damage initiation

Von Mises yield criterion is used for micro-damage initiation. When von Mises stress of an element exceeds a threshold, it is assumed that micro-damage is initiated in the element. The von Mises stress is calculated using Eq. (3.3). Damage variable D is formulated as

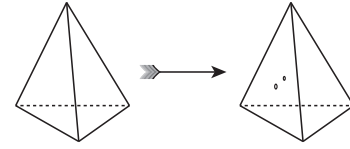


Figure 3.14: Step1

$$D = \begin{cases} 0 & (\sigma_{Mises} < \sigma_t) \\ V_{\text{void}}/V_e & (\sigma_{Mises} \geq \sigma_t) \end{cases} \quad (3.32)$$

where σ_{Mises} is the von Mises stress of an element and σ_t is the stress threshold for micro-damage initiation.

Step2: Micro-damage growth

In an element, the sum volume of micro-damage, such as voids and cracks, is V_{void} . In the proposed damage model, it is assumed that the volume change rate of V_{void} is in

proportion to volume transformation of an element ΔV_e . Therefore, $\dot{V}_{\text{void}}$ is formulated by

$$\dot{V}_{\text{void}} = \lambda \Delta V_e, \quad (3.33)$$

where λ is micro-damage growth coefficient.

Damage variable D is formulated as

$$D = V_{\text{void}}/V_e, \quad (3.34)$$

where V_e is the volume of the element. Volume strain ϵ_v is defined as

$$\epsilon_v = \Delta V_e/V_e. \quad (3.35)$$

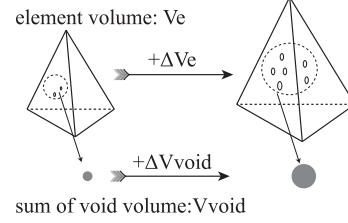


Figure 3.15: Step2

The substitution of Eq. (3.33) and Eq. (3.35) into Eq. (3.34), and solving for $\dot{D}$ gives

$$\dot{D} = \lambda \epsilon_v \quad (3.36)$$

The damage is non-invertible. Therefore, Eq. (3.36) is a conditional equation. In the time increment $[t_{(n-1)}, t_{(n)}]$, the progression of D is limited by:

- (i) $\epsilon_{v(n-1)} \cdot \epsilon_{v(n)} < 0$
- (ii) $\epsilon_{v(n-1)} \cdot \epsilon_{v(n)} \geq 0$ and $|\epsilon_{v(n)}| > |\epsilon_{v(n-1)}|$

The stiffness loss increases as V_{void} growth. The damaged stiffness matrix $\mathbf{K}_e(\omega)$ of an element after stiffness loss is given by

$$\mathbf{K}_e(\omega) = (V_e - V_{\text{void}}) (\mathbf{B}_e)^T \mathbf{C}_e \mathbf{B}_e = (1 - D)V_e (\mathbf{B}_e)^T \mathbf{C}_e \mathbf{B}_e = \omega \mathbf{K}_e. \quad (3.37)$$

Step3: Coalescence of micro-damages

When $D = 1$, it is assumed that the element is completely fractured, and the stiffness of an element is set to zero as

$$\mathbf{K}_e = \mathbf{0} \quad (3.38)$$

Hence, the element is removed to express fracture.

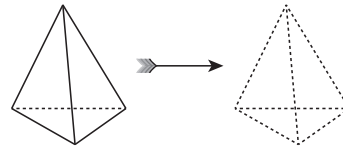


Figure 3.16: Step3

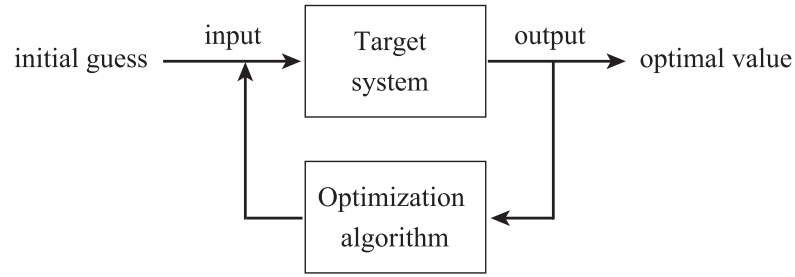


Figure 3.17: Image of optimization algorithm.

3.4.2 Parameter identification for damage fracture model A

To perform a reality simulation by using the proposed damage model for brain parenchyma, mechanical properties of brain parenchyma and parameters of the proposed damage model should be identified. The properties and parameters to be identified are Young's modulus E , Rayleigh damping scalars a and b , threshold of von Mises stress for micro-damage initiation σ_t , and micro-damage growth rate coefficient λ . These properties and parameters were identified by solving an optimization problem.

An optimization problem is the problem of finding the best solution from all feasible solutions. In the case of optimization to the target value, the minimal difference from the target value is the best solution. In the search of solutions by using the optimization algorithm, a target system is first required. When the target system is provided with an input value, it needs a mechanism to return some output value, for example, actual measurement experiments. For such a system, the optimization algorithm provides an input value and judges whether the response output from the system is optimal. If not, the optimization algorithm repeats the process until an optimal solution is obtained. Fig. 3.17 describes the image of the searching solution using the optimization algorithm.

Some earlier studies use the optimization problem to identify mechanical properties of materials [80, 81]. In the optimization problem, a fitted curve of experimental data of tensile tests at 0.1 (mm/s) was used to identify properties and parameters. The simulation was performed with the same conditions as those of the tensile tests. A variety of the properties and parameters were used to determine the minimal error between simulation result and fitted curve. The error was calculated in the simulation and properties, and parameters were decided on the optimization problem solver.

In this study, *fmincon()* function in the MATLAB Optimization ToolboxTM was

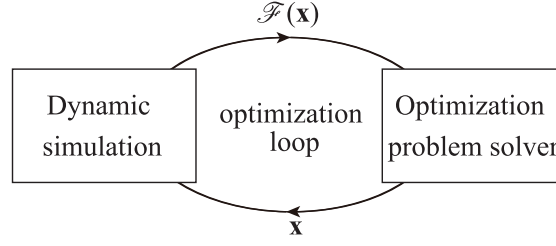


Figure 3.18: Identification process.

used as the solver. $fmincon()$ is a nonlinear constrained optimization solver using sequential quadratic programming (SQP). The dynamic simulator was programmed in C++. The $fmincon()$ function of MATLAB calls the dynamic simulator as a child process. Fig. 3.18 illustrates the identification process.

The optimization problem is defined as follows:

(i) Parameter vector

$$\mathbf{x} = [E \quad \sigma_t \quad a \quad b \quad \gamma] \quad (3.39)$$

(ii) Objective function

minimize $f(\mathbf{x})$

$$\mathcal{F}(\mathbf{x}) = w_1 \sum_{\varepsilon=0.0}^{0.2} \sigma_{\text{err}}^2(\varepsilon) + w_2 \sum_{\varepsilon=0.2}^{0.7} \sigma_{\text{err}}^2(\varepsilon) + w_3 \sum_{\varepsilon=0.7}^{0.9} \sigma_{\text{err}}^2(\varepsilon) \quad (3.40)$$

$$\sigma_{\text{err}}(\varepsilon) = \sigma_{\text{fit}}(\varepsilon) - \sigma_{\text{sim}}(\varepsilon)$$

where w_i ($i = 1, 2, 3$) are weight coefficients. $\sigma_{\text{fit}}(\varepsilon)$ is the stress of the fitted curve. $\sigma_{\text{sim}}(\varepsilon)$ is the stress of the simulation result.

(iii) Range of parameters are subject to

$$\begin{aligned} E &\in [10^3, 10^4], & \sigma_t &\in [10, 10^4], \\ \alpha, \beta &\in [0, 10], & \lambda &\in [10^{-4}, 1] \end{aligned} \quad (3.41)$$

(iv) Initial parameters

$$\mathbf{x}_0 = [3000 \quad 1000 \quad 1 \quad 1 \quad 0.03] \quad (3.42)$$

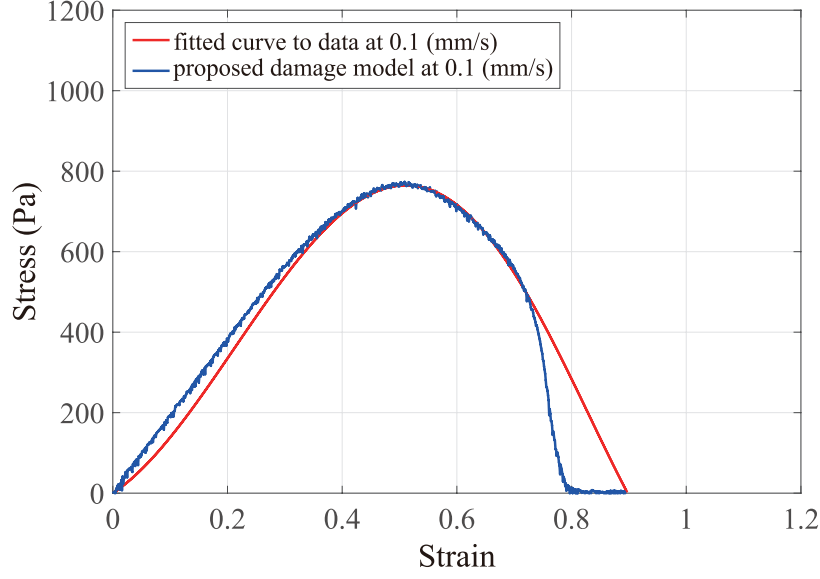


Figure 3.19: Stress–strain curve obtained through a dynamic simulation with identified parameters and damage fracture model A, and the curve obtained by fitting to experimental results with tensile velocity 0.1 mm/s.

(v) Optimized result

$$\mathbf{x}_{\text{opt}} = \begin{bmatrix} 3000 & 1000 & 1.844 & 1.127 & 0.0307 \end{bmatrix} \quad (3.43)$$

The weight coefficients were set to $w_1 = w_3 = 0.1$, and $w_2 = 0.8$, so that the error in the intermediate range of a stress–strain curve is minimized.

The stress-strain curve obtained by dynamic simulation with $\mathbf{x}_{\text{opt}}$ and the curve obtained by fitting to experimental results with tensile velocity of 0.1 (mm/s) are plotted in Fig. 3.19. As shown in Fig. 3.19, identification of parameters through the optimization problem is useful.

3.5 Damage Fracture Model B [82]

This section presents the proposed damage fracture model B based on CDM. Moreover, the parameters required for damage fracture model B are identified.

3.5.1 Damage fracture model B

The fracture process is also expressed using the following three steps: (1) micro-damage initiation, (2) micro-damage growth, and (3) coalescence of micro-damage, as illustrated in Figs. 3.13 – 3.16. These steps are summarized as follows:

Step1: Micro-damage initiation

The von Mises yield criterion is used for micro-damage initiation. The damage variable D is formulated using Eq. (3.34).

Step2: Micro-damage growth

In this damage model, it is assumed that the volume of micro-damage V_{void} grows because of the increase in von Mises stress. The evolution equation for the void volume fraction is formulated as

$$\dot{D} = \begin{cases} \lambda(1-D)(\sigma_{\text{Mises}} - \sigma_t) & (\sigma_{\text{Mises}} \geq \sigma_t), \\ 0 & (\sigma_{\text{Mises}} < \sigma_t), \end{cases} \quad (3.44)$$

where λ is a coefficient.

The stiffness loss increases with V_{void} growth. The damaged stiffness matrix $\mathbf{K}_e(\omega)$ of an element after stiffness loss is given by Eq. (3.37).

Step3: Coalescence of micro-damages

When the void volume increases sufficiently, the voids in the RVE coalesce with voids in adjacent RVEs. Therefore, it is assumed that if the integrity ω of an RVE reduces more than a threshold ω_t , coalescence of microscopic voids evolves rapidly into fracture. Hence, the element is removed to express fracture. In this study, ω_t was set to 0.005.

3.5.2 Parameter identification for damage fracture model B

Young's modulus E , stress threshold σ_t for micro-damage initiation, Rayleigh damping scalars a , b , and void volume growth rate coefficient γ were identified by solving

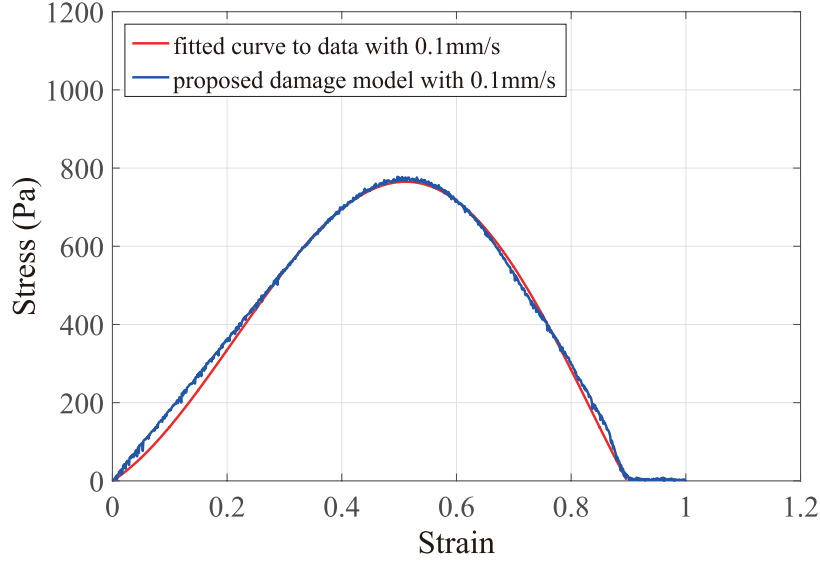


Figure 3.20: Stress–strain curve obtained through dynamic simulation using the identified parameters.

an optimization problem. Tensile test simulations were performed in the optimization loop with a brain parenchyma model illustrated in Fig. 3.7 at a tensile velocity of 0.1 mm/s. The parameter vector $\mathbf{x}$ was provided from an optimization problem solver to the dynamic simulator. The dynamic simulator performs a tensile test with the given parameter vector and returns the objective function $\mathcal{F}(\mathbf{x})$ to the optimization problem solver. This process is iterated until $\mathcal{F}(\mathbf{x})$ is minimized. The objective function $\mathcal{F}(\mathbf{x})$ is defined as follows:

$$\mathcal{F}(\mathbf{x}) = \sum_{\epsilon=0.0}^{0.9} (\sigma_{\text{fit}}(\epsilon) - \sigma_{\text{sim}}(\epsilon))^2, \quad (3.45)$$

$$\mathbf{x} = \begin{bmatrix} E & \sigma_t & a & b & \gamma \end{bmatrix}, \quad (3.46)$$

where $\mathbf{x}$ is a parameter vector, $\sigma_{\text{fit}}(\epsilon)$ is the stress given by the fitted curve at the tensile velocity 0.1 mm/s (the solid line in Fig. 3.5), and $\sigma_{\text{sim}}(\epsilon)$ is the stress obtained from tensile test simulations.

The *fmincon()* function was used to solve the optimization problem, which is defined as follows:

$$\text{minimize} \quad \mathcal{F}(\mathbf{x}),$$

$$\text{subject to} \quad E \in [1.0 \times 10^3, 5.0 \times 10^3], \quad \sigma_t \in [1.0 \times 10^2, 2.5 \times 10^3], \\ a, b \in [1.0 \times 10^{-4}, 10.0], \quad \gamma \in [1.0 \times 10^{-7}, 5.0 \times 10^{-6}].$$

The initial guess was set to

$$\mathbf{x}_0 = [2500 \quad 800 \quad 0.6 \quad 0.6 \quad 3.0 \times 10^{-6}].$$

The optimized parameters were

$$\mathbf{x}_{opt} = [2823.8 \quad 1183.6 \quad 1.174 \quad 0.958 \quad 3.316 \times 10^{-6}].$$

The stress-strain curves obtained through dynamic simulation with $\mathbf{x}_{opt}$ and curves obtained by fitting to experimental results with a tensile velocity of 0.1 mm/s are plotted in Fig. 3.20. As shown, the error between the simulation and experimental results (fitted curve of the experimental data with tensile velocity of 0.1 (mm/s)) is small. Therefore, the usefulness of the parameter identification is proved by solving the optimization problem.

3.6 Tensile Simulation

In Sections 3.4.1 and 3.5.1, damage fracture models A and B were described. Moreover, in Sections 3.4.2 and 3.5.2, parameters of the proposed damage fracture models A and B were identified. This section describes the verification of the models and parameters. Tensile simulations using the proposed models and identified parameters are run. Verification is conducted by comparing the results of simulations and tensile tests. Fig. 3.21 shows the flow chart of the numerical calculation progress using the damage fracture models.

The simulation settings are described as follows.

- Simulation model:

The size of model used in the simulation is the average size of specimens, as shown in Table 3.1. Fig. 3.7 illustrates the model.

- Contact nodes:

Nodes of the upper and bottom surfaces are $z = 15$ and $z = 0$ (in Fig. 3.7).

- Forced displacement of contact nodes:

Upper surface nodes are described as

$$u_x = u_y = u_z = 0,$$

and bottom surface nodes is

$$\begin{aligned} u_x = u_y = 0, u_z = -0.1t \quad & \text{(tensile velocity is 0.1(mm/s))}, \\ \text{or} \quad u_x = u_y = 0, u_z = -1.0t \quad & \text{(tensile velocity is 1.0(mm/s))}. \end{aligned}$$

- Parameters: mechanical properties and parameters of the proposed damage fracture models are summarized in Table 3.4.

Fig. 3.22 illustrates the sequences of the tensile simulations. The simulation result was compared with the experimental results (Fig. 3.4). Stress–strain curves of tensile simulations with tensile velocities of 0.1 (mm/s) and 1.0 (mm/s) using the model A are plotted in Fig. 3.23, and the simulations using the model B are plotted in Fig. 3.24.

Figures 3.23 and 3.24 show the comparisons between the simulation and experimental results. As parameters were identified using the experimental data of tensile velocities of 0.1 (mm/s), as shown in Figs. 3.23 (a) and 3.24 (a), the simulation results at a tensile velocity of 0.1 (mm/s) fit the corresponding experimental results.

To verify the proposed damage models and identified properties and parameters, tensile simulations were performed using identified parameters at the tensile velocity of 1.0 (mm/s). As shown in Figs. 3.23 (b) and 3.24 (b), although parameters are identified using experimental results with tensile velocity of 0.1 (mm/s), the damage and fracture model satisfactorily reproduces the experimental results with tensile velocity 1.0 (mm/s).

Animation pictures of the simulation (Fig. 3.22) show the fracture process of brain parenchyma. The fracture develops gradually with the deformation in the longitudinal direction. These results validate the effectiveness of the proposed damage fracture models. The comparison of the results of models A and B shows that model B is better than model A regarding accuracy, especially during a large strain.

Table 3.4: Mechanical properties and parameters of proposed damage fracture models A and B for tensile simulations.

Parameters	Model A	Model B
Number of nodes	996	
Number of tetrahedral elements	3779	
Density (kg/m ³)	1036	
Poisson's ratio	0.4925	
Sampling time (ms), at 0.1 (mm/s)	150,	150,
at 1.0 (mm/s)	15	20
Young's modulus (Pa)	3000	2823.8
Stress threshold (Pa)	$\sigma_t = 1000$	$\sigma_t = 1183.6$
Rayleigh damping scalars	$a = 1.844,$ $b = 1.127$	$a = 1.174,$ $b = 0.958$
Void volume growth rate coefficient	$\lambda = 3.07 \times 10^{-2}$	$\lambda = 3.316 \times 10^{-6}$
integrity threshold	$\omega_t = 0.0$	$\omega_t = 0.005$

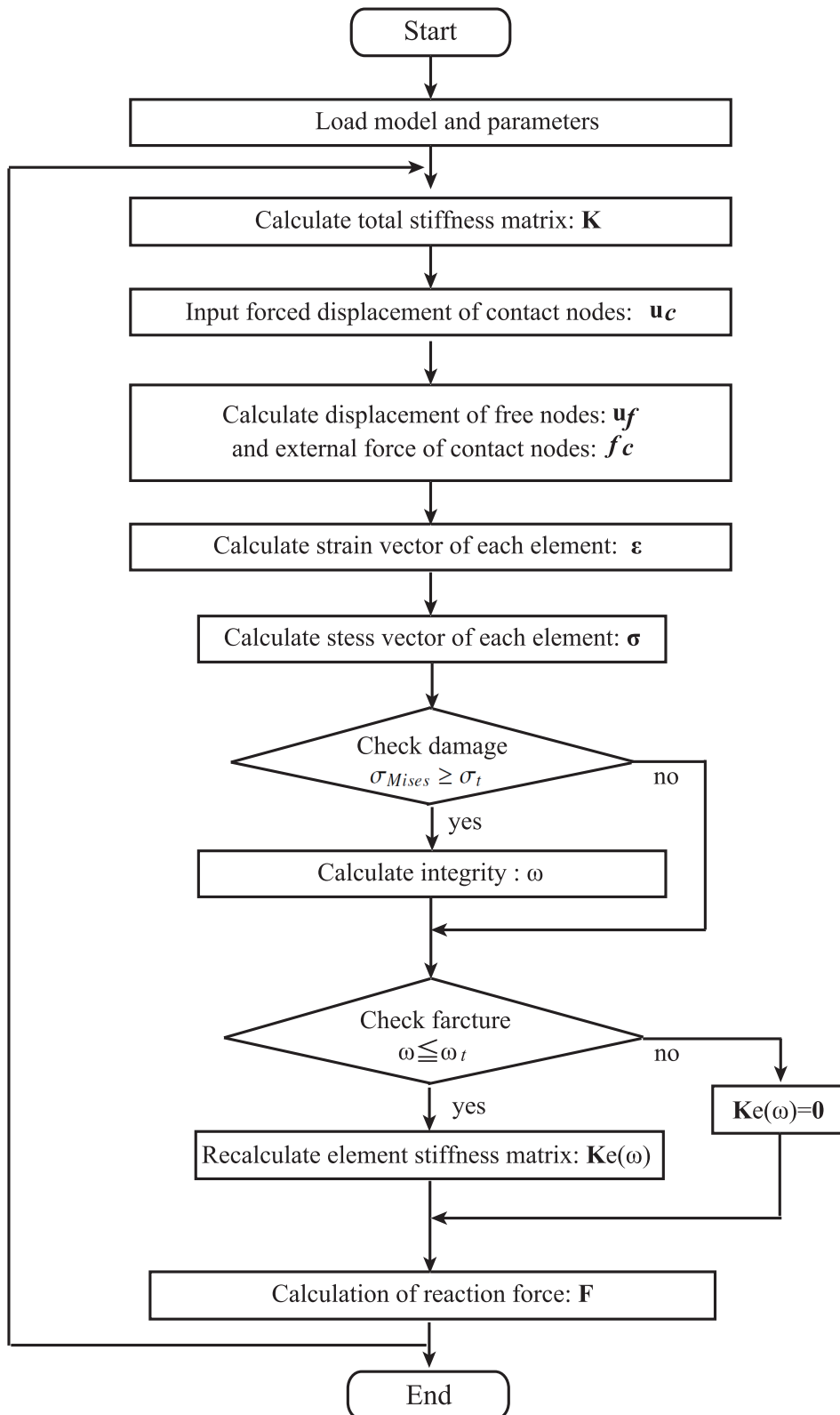
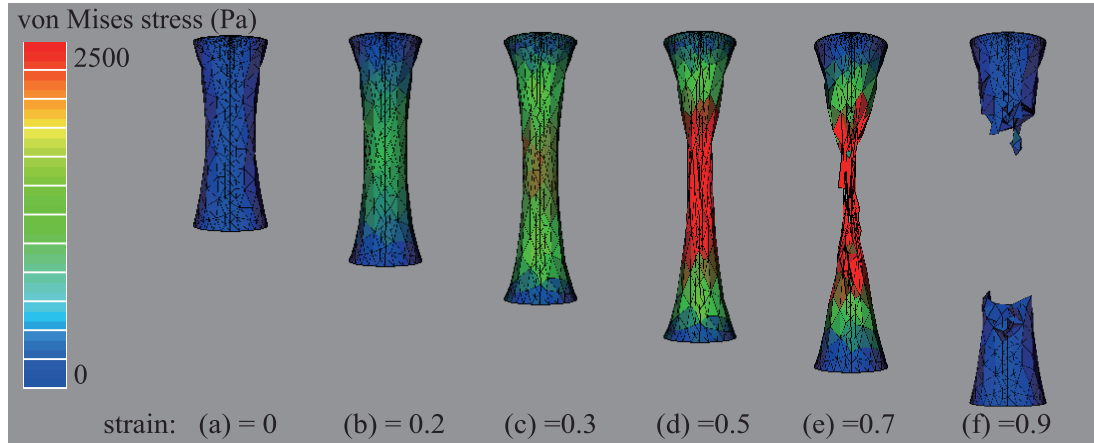
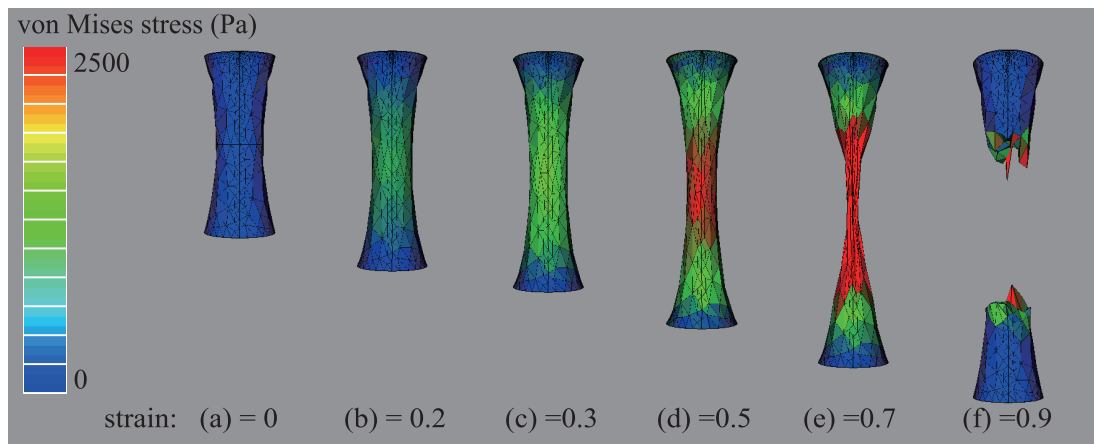


Figure 3.21: Flow chart of the numerical calculation progress with damage fracture model

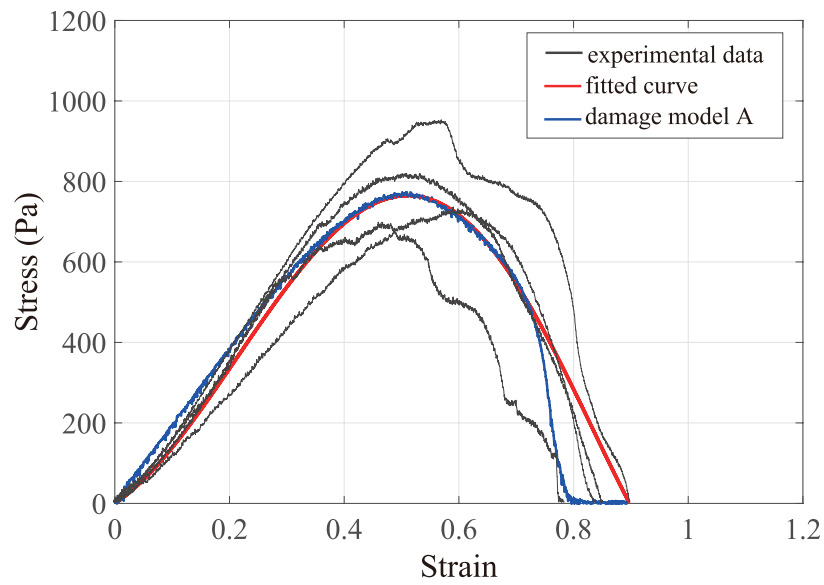


(a) Model A, 1.0 (mm/s).

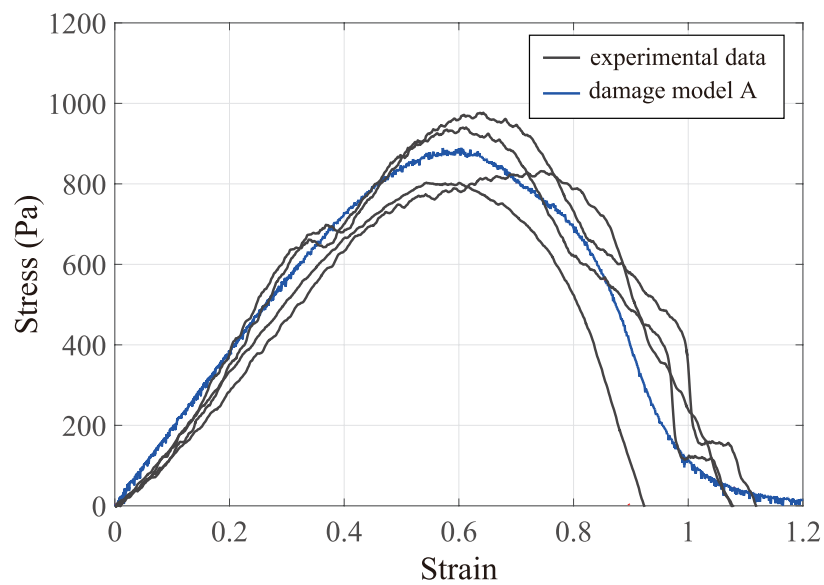


(b) Model B, 0.1 (mm/s).

Figure 3.22: Sequences of the simulation at a tensile velocity of 0.1 mm/s with damage fracture models A and B.

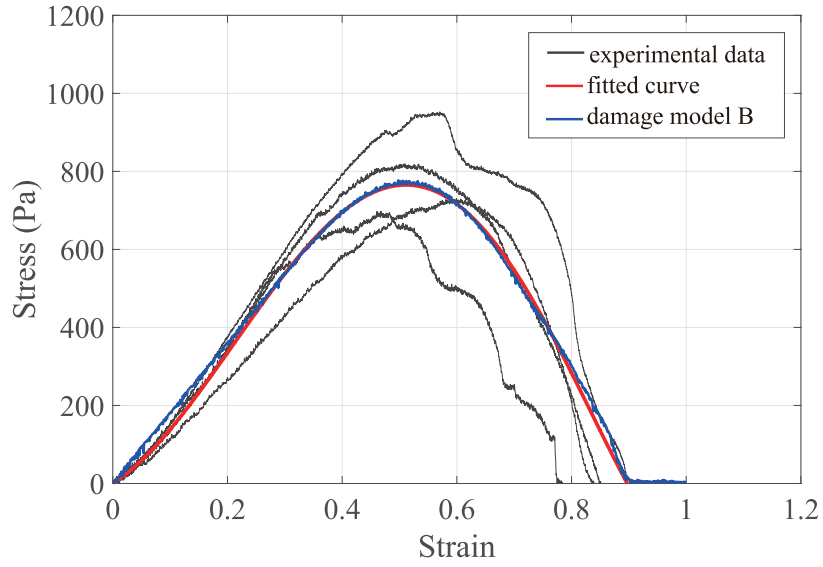


(a) Tensile velocity: 0.1 (mm/s)

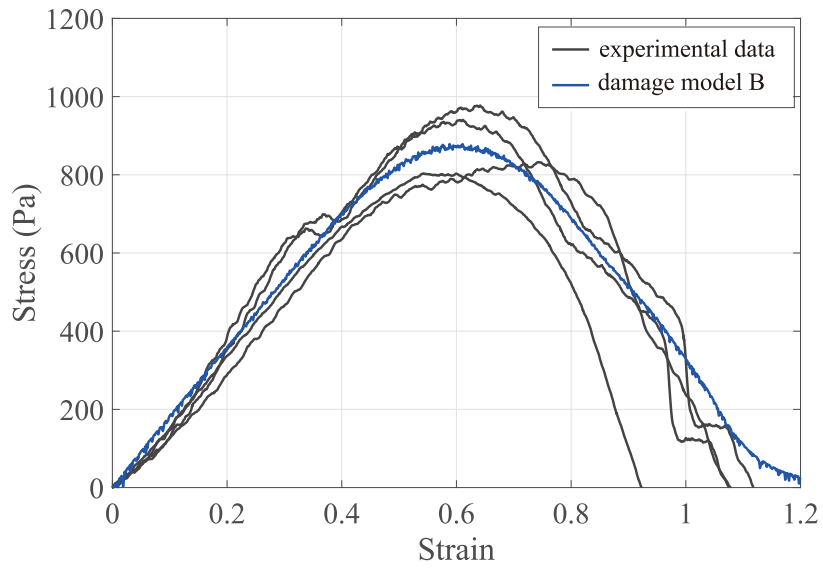


(b) Tensile velocity: 1.0 (mm/s)

Figure 3.23: Stress—strain curves of experiments and the simulation using proposed damage model A.



(a) Tensile velocity: 0.1 (mm/s)



(b) Tensile velocity: 1.0 (mm/s)

Figure 3.24: Stress–strain curves of experiments and the simulation using proposed damage model B.

3.7 Summary

The details of tensile tests and tensile simulations were presented in this chapter.

When a specimen of brain tissue is stretched, it generally shows ductile fractures. However, conventional fracture or damage models cannot accurately reproduce ductile fracture characteristics. To simulate dissection and removal operations in brain surgery, two simple damage fracture models for brain parenchyma were proposed.

The novelty of this research consists of the following four points:

1. Fracture characteristics of brain parenchyma were verified experimentally by performing tensile tests using porcine brain parenchyma at two low tensile velocities (0.1 mm/s and 1.0 mm/s).
2. Two simple damage fracture models were proposed and verified by comparing experimental and simulation results.
3. Mechanical properties and proposed model parameters were identified.

Chapter 4. Stress Relaxation Model

4.1 Introduction

In this chapter, the details of stress relaxation tests model are presented. The main goal of this research is to develop a stress relaxation model of brain parenchyma to simulate the brain retraction for haptic brain surgery simulations.

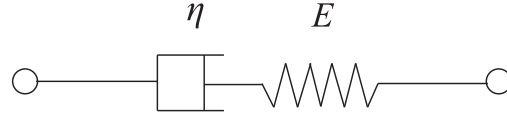
Brain retraction is a very important surgical technique used for lesion exposure in neurosurgeries. However, excessive load of brain retraction may cause permanent damage to the nerve. Therefore, development of surgical simulators, which can accurately simulate the stress and force of brain reaction, is required. To simulate brain retraction, a stress relaxation model was proposed.

This chapter is organized as follows. Section 4.2 presents a previous study of generalized Maxwell model using FE method. The computational approach of the model is described, and its disadvantage is clearly articulated.

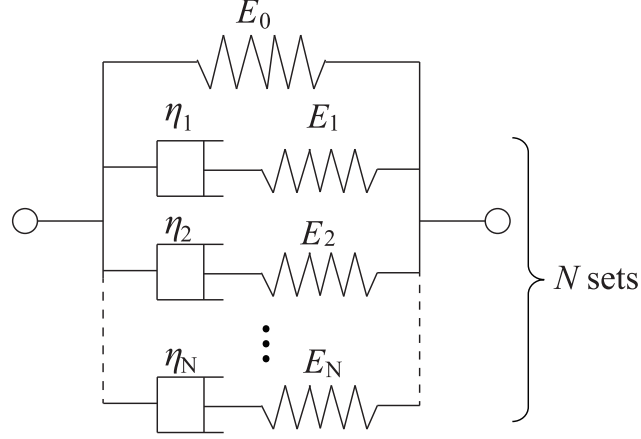
In Section 4.3, a stress relaxation model of brain parenchyma based on the generalized Maxwell model but also considering the inertial term is presented.

Section 4.4 describes the stress relaxation tests. The stress relaxation characteristics were measured at strains of 0.1, 0.2, and 0.3. The strains were applied using strain rates of $0.01 \text{ (s}^{-1}\text{)}$ and $0.1 \text{ (s}^{-1}\text{)}$. Parameters of the proposed model were identified by solving an optimization problem. Stress relaxation simulations were performed using the model and identified parameters. Moreover, the model is verified to simulate the stress relaxation characteristics of brain parenchyma.

In Section 4.6, brain retraction experiments were performed to verify mechanical characteristics by using porcine brains. The retraction velocities were 1 and 10 (mm/s). Parameters of the proposed model were identified using the retraction experimental results. A 3D simulation model was built from MRI data of a porcine brain. Retraction simulations were conducted, and it is verified that brain retraction can be simulated



(a) Maxwell model



(b) Generalized Maxwell model

Figure 4.1: Schematic of Maxwell and generalized Maxwell models

using the proposed model.

4.2 Previous Studies on Viscoelasticity Model

In this section, FEM implementation of the generalized Maxwell model [84] is introduced. The generalized Maxwell model (Fig. 4.1 (b)) is interpreted as a linear spring in parallel to N sets of a Maxwell element (Fig. 4.1 (a)). Elasticity E of the model is a time depending modulus and is defined by the Prony series expression as

$$E(t) = E_0 + \sum_{i=1}^N E_i e^{-t/\tau_i}, \quad (4.1)$$

where $\tau_i = \eta_i/E_i$ is called the relaxation time.

The stress σ at time t is the integral over the starting time s ,

$$\begin{aligned}\sigma(t) &= \int_0^t E(t-s) \frac{\partial \varepsilon}{\partial s} ds \\ &= E_0 \varepsilon(t) + \sum_{i=1}^N \int_0^t E_i e^{-\frac{t-s}{\tau_i}} \frac{\partial \varepsilon}{\partial s} ds \\ &= \sigma_0(t) + \sum_{i=1}^N h_i(t),\end{aligned}\tag{4.2}$$

where ε is the strain. When strain ε remains constant, the stress relaxes and converges to $\sigma(\infty) = \sigma_0 = E_0 \varepsilon$.

Equation (4.2) is discretized as follows:

$$\sigma^{(n+1)} = \sigma_0^{(n+1)} + \sum_{i=1}^N h_i^{(n+1)},\tag{4.3}$$

where

$$h_i^{(n+1)} = e^{-\Delta t / \tau_i} h_i^{(n)} + r_i \frac{1 - e^{-\Delta t / \tau_i}}{\Delta t / \tau_i} \left[\sigma_0^{(n+1)} - \sigma_0^{(n)} \right],\tag{4.4}$$

$$r_i = \frac{E_i}{E_0},\tag{4.5}$$

and Δt is the sampling time.

In the linear FEM, strain vector ε_e and stress vector σ_e of an element are calculated by Eq. (2.7) and (2.10). In the FEM formulation, the elastic contribution term of Eq. (4.3) becomes

$$(\sigma_e)_0^{(n+1)} = \mathbf{D}_e \varepsilon_e^{(n+1)} = \mathbf{D}_e \mathbf{B}_e \mathbf{u}_e^{(n+1)}.\tag{4.6}$$

Therefore, FEM formulation of Eq. (4.3) is given as follows:

$$\begin{aligned}\sigma_e^{(n+1)} &= \mathbf{D}_e \mathbf{B}_e \mathbf{u}_e^{(n+1)} + \sum_{i=1}^N \left[e^{-\Delta t / \tau_i} \mathbf{h}_i^{(n)} + r_i A_i \mathbf{D}_e \mathbf{B}_e \left(\mathbf{u}_e^{(n+1)} - \mathbf{u}_e^{(n)} \right) \right] \\ &= \left(1 + \sum_{i=1}^N r_i A_i \right) \mathbf{D}_e \mathbf{B}_e \mathbf{u}_e^{(n+1)} + \sum_{i=1}^N e^{-\Delta t / \tau_i} \mathbf{h}_i^{(n)} - \left(\sum_{i=1}^N r_i A_i \right) \mathbf{D}_e \mathbf{B}_e \mathbf{u}_e^{(n)}\end{aligned}\tag{4.7}$$

where

$$A_i = \frac{1 - e^{-\Delta t / \tau_i}}{\Delta t / \tau_i}.\tag{4.8}$$

$\mathbf{h}_i$ is updated as

$$\mathbf{h}_i^{(n+1)} = e^{-\Delta t / \tau_i} \mathbf{h}_i^{(n)} + r_i A_i \mathbf{D}_e \mathbf{B}_e \left(\mathbf{u}_e^{(n+1)} - \mathbf{u}_e^{(n)} \right).\tag{4.9}$$

The element force vector $\mathbf{f}_e$ and element stiffness matrix $\mathbf{K}_e$ in static FEM are given by

$$\mathbf{f}_e = \int \mathbf{B}_e^T \boldsymbol{\sigma}_e dV = \mathbf{K}_e \mathbf{u}_e, \quad (4.10)$$

$$\mathbf{K}_e = V_e \mathbf{B}_e^T \mathbf{D}_e \mathbf{B}_e, \quad (4.11)$$

where V_e is the element volume.

By using Eqs. (4.7), (4.10), and (4.11), the elemental force vector is calculated as follows:

$$\mathbf{f}_e^{(n+1)} = (\mathbf{K}_e)_n \mathbf{u}_e^{(n+1)} + (\mathbf{f}_e)_{\text{hist}}^{(n+1)}, \quad (4.12)$$

where

$$(\mathbf{K}_e)_n = \left(1 + \sum_{i=1}^N r_i A_i \right) \mathbf{K}_e, \quad (4.13)$$

$$(\mathbf{f}_e)_{\text{hist}}^{(n+1)} = \mathbf{h}_{\text{hist}}^{(n+1)} - (\mathbf{K}_e)_{\text{hist}} \mathbf{u}_e^{(n)}, \quad (4.14)$$

$$\mathbf{h}_{\text{hist}}^{(n+1)} = (V_e \mathbf{B}_e^T) \sum e^{-\Delta t / \tau_i} \mathbf{h}_i^{(n)}, \quad (4.15)$$

$$(\mathbf{K}_e)_{\text{hist}} = \left(\sum_{i=1}^N r_i A_i \right) \mathbf{K}_e. \quad (4.16)$$

By extending Eq. (4.12) to include the whole simulation model, it is rewritten as

$$\mathbf{f}^{(n+1)} = \mathbf{K}_n \mathbf{u}^{(n+1)} + \mathbf{f}_{\text{hist}}^{(n+1)}, \quad (4.17)$$

where

$$\mathbf{K}_n = \mathbf{A}_{e=1}^{\text{all}} (\mathbf{K}_e)_n, \quad (4.18)$$

$$\mathbf{f}_{\text{hist}} = \mathbf{A}_{e=1}^{\text{all}} (\mathbf{f}_e)_{\text{hist}}, \quad (4.19)$$

$\mathbf{A}_{e=1}^{\text{all}}$ is a finite element assembly operator. $\mathbf{f}$ and $\mathbf{u}$ are force and displacement vectors of all nodes, respectively.

4.3 Numerical model of viscoelasticity with an inertial term [85]

A generalized Maxwell model using FE method was formulated using the FEM as Eq. (4.17) [86, 84]. There is no inertial term in Eq. (4.17). As mentioned in Section 4.2, the computation will be more feasible by adding an inertial term. Therefore, a viscoelastic model with an inertial term is proposed as follows:

$$\mathbf{f}^{(n+1)} = \mathbf{K}_n \mathbf{u}^{(n+1)} + \mathbf{f}_{\text{hist}}^{(n+1)} + \mathbf{M} \ddot{\mathbf{u}}^{(n+1)}, \quad (4.20)$$

where $\mathbf{M}$ is mass matrix and $\ddot{\mathbf{u}}$ is acceleration vector of nodes. $\mathbf{M}$ is calculated using Eq. (2.23), and $\ddot{\mathbf{u}}$ is updated using Eq. (2.31).

In this study, force and displacement for contact and free nodes are calculated separately (represented in Section 2.3.3). Therefore, Eq. (4.20) is divided as follows:

$$\begin{bmatrix} \mathbf{f}_c^{(n+1)} \\ \mathbf{f}_f^{(n+1)} \end{bmatrix} = \begin{bmatrix} (\mathbf{K}_{cc})_n & (\mathbf{K}_{cf})_n \\ (\mathbf{K}_{fc})_n & (\mathbf{K}_{ff})_n \end{bmatrix} \begin{bmatrix} \mathbf{u}_c^{(n+1)} \\ \mathbf{u}_f^{(n+1)} \end{bmatrix} + \begin{bmatrix} (\mathbf{f}_c)_{\text{hist}}^{(n+1)} \\ (\mathbf{f}_f)_{\text{hist}}^{(n+1)} \end{bmatrix} + \begin{bmatrix} \mathbf{M}_c & \mathbf{0} \\ \mathbf{0} & \mathbf{M}_f \end{bmatrix} \begin{bmatrix} \ddot{\mathbf{u}}_c^{(n+1)} \\ \ddot{\mathbf{u}}_f^{(n+1)} \end{bmatrix}, \quad (4.21)$$

where subscript “c” expresses contact-node-related terms, and subscript “f” expresses free-node-related terms.

As the contact nodes move with the surgical instrument operated by a surgeon, $\mathbf{u}_c^{(n+1)}$ is known, while $\mathbf{f}_f^{(n+1)} = 0$. By substituting Eq. (2.31) into Eq. (4.21), the displacement of free nodes $\mathbf{u}_f^{(n+1)}$ and external force of contact node $\mathbf{f}_c^{(n+1)}$ are calculated separately as follows:

$$\begin{aligned} \mathbf{u}_f^{(n+1)} = & - \left[(\mathbf{K}_{ff})_n + a_3 \mathbf{M}_f \right]^{-1} \left[(\mathbf{K}_{fc})_n \mathbf{u}_c^{(n+1)} \right. \\ & \left. + (\mathbf{f}_f)_{\text{hist}}^{(n+1)} + \mathbf{M}_f \left(a_3 \mathbf{u}_f^{(n)} + a_4 \dot{\mathbf{u}}_f^{(n)} + a_5 \ddot{\mathbf{u}}_f^{(n)} \right) \right], \end{aligned} \quad (4.22)$$

$$\begin{aligned} \mathbf{f}_c^{(n+1)} = & ((\mathbf{K}_{cc})_n + a_3 \mathbf{M}_c) \mathbf{u}_c^{(n+1)} + (\mathbf{K}_{fc})_n \mathbf{u}_f^{(n+1)} + (\mathbf{f}_c)_{\text{hist}}^{(n+1)} \\ & - \mathbf{M}_c \left(a_3 \mathbf{u}_c^{(n)} + a_4 \dot{\mathbf{u}}_c^{(n)} + a_5 \ddot{\mathbf{u}}_c^{(n)} \right). \end{aligned} \quad (4.23)$$

The displacement vector $\mathbf{u}_f^{(n+1)}$ of free nodes is calculated using Eq. (4.22). It is easily understood that if mass matrix $\mathbf{M}_f$ is not included, the calculation of Eq. (4.22)

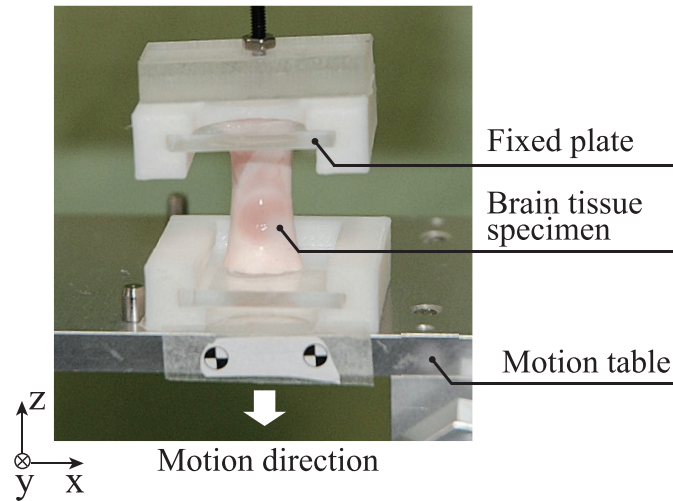


Figure 4.2: Measurement equipment of stress relaxation tests with porcine brain parenchyma

becomes unstable when $(\mathbf{K}_{ff})_n$ is singular. By considering inertial term with a diagonal mass matrix $\mathbf{M}$, the calculation Eq. (4.22) is always feasible and stable. This is the main advantage of including the inertial term.

4.4 Stress Relaxation Tests [85]

4.4.1 Stress relaxation tests

Two heads of pigs, which were killed on the experiment day, were collected. Seven specimens were taken from different areas of the two cerebra. Specimens were cut into cylindrical shape by using a cork borer. The diameter of the specimens is 5 ± 1 (mm) and height is 10 ± 1 (mm). White and gray matters were mixed in the specimens. Fig. 4.2 illustrates the specimens and measurement equipment.

Stress relaxation tests were conducted, and the specimens were pulled at a constant strain rate to a certain strains of 0.1, 0.2, and 0.3. After each strain was reached, specimens were held in their pure form for several dozens of seconds. Four successful specimens were obtained after the removal of outlier results. In the four specimens, one specimen was pulled at a strain rate of $0.1 \text{ (s}^{-1}\text{)}$, and the other three specimens were pulled at a strain rate of $0.01 \text{ (s}^{-1}\text{)}$.

Fig. 4.3 illustrates a stress relaxation test result. of specimen No.6 pulled at strain

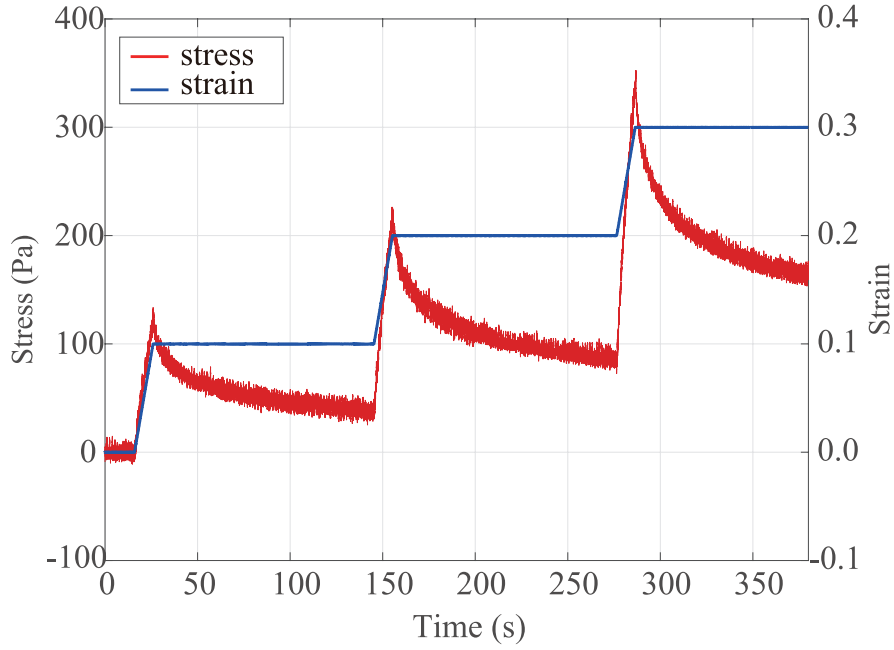


Figure 4.3: Example (specimen No.6) of stress relaxation test results at a strain rate of $0.01 \text{ (s}^{-1}\text{)}$.

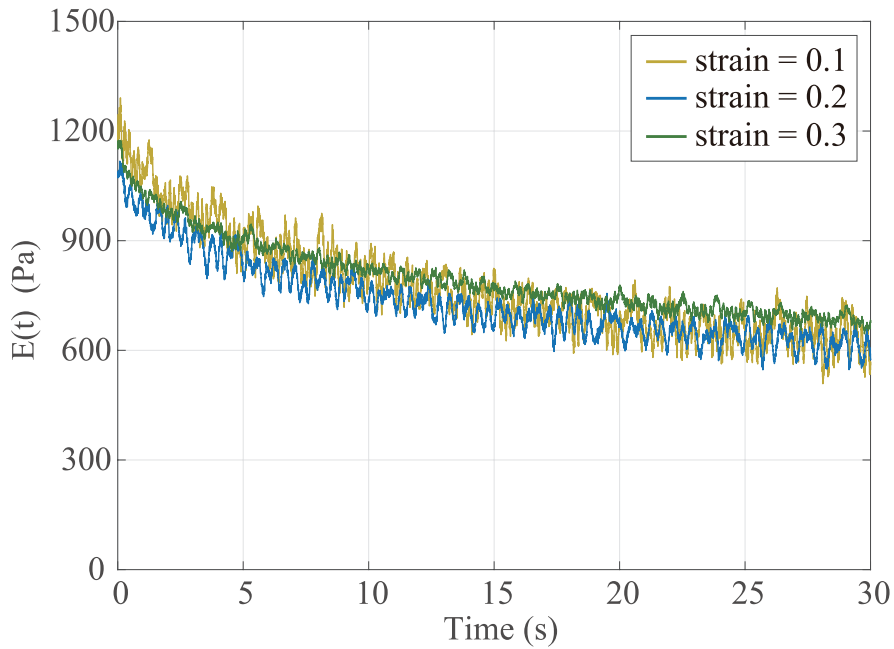


Figure 4.4: Example (specimen No.6) of $E(t)$ calculated by Fig. 4.3 at strains of 0.1, 0.2, and 0.3.

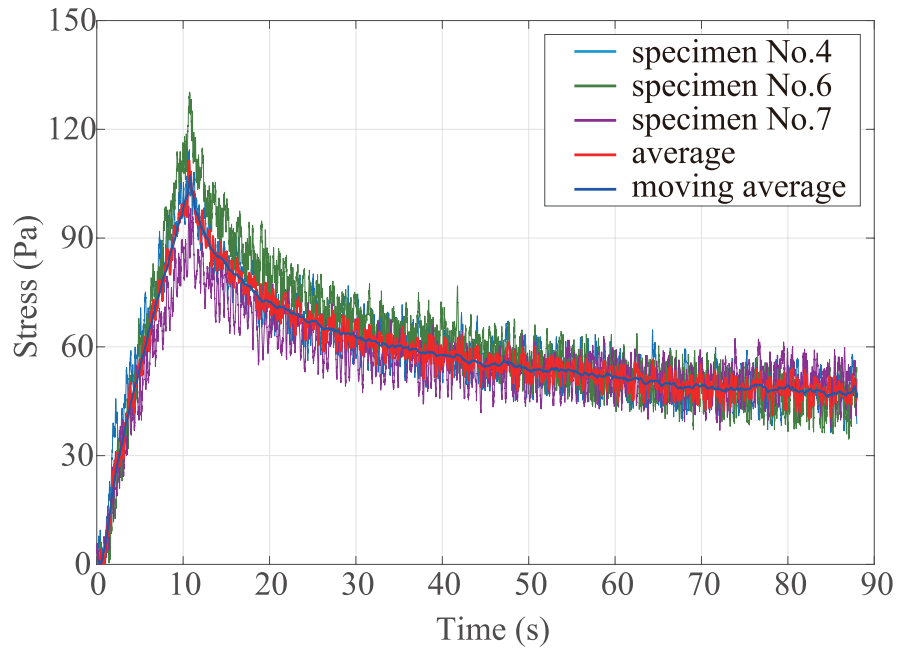


Figure 4.5: Three experimental results of tensile and stress relaxation tests with a strain rate of $0.01 \text{ (s}^{-1}\text{)}$ from strain = 0 to strain = 0.1.

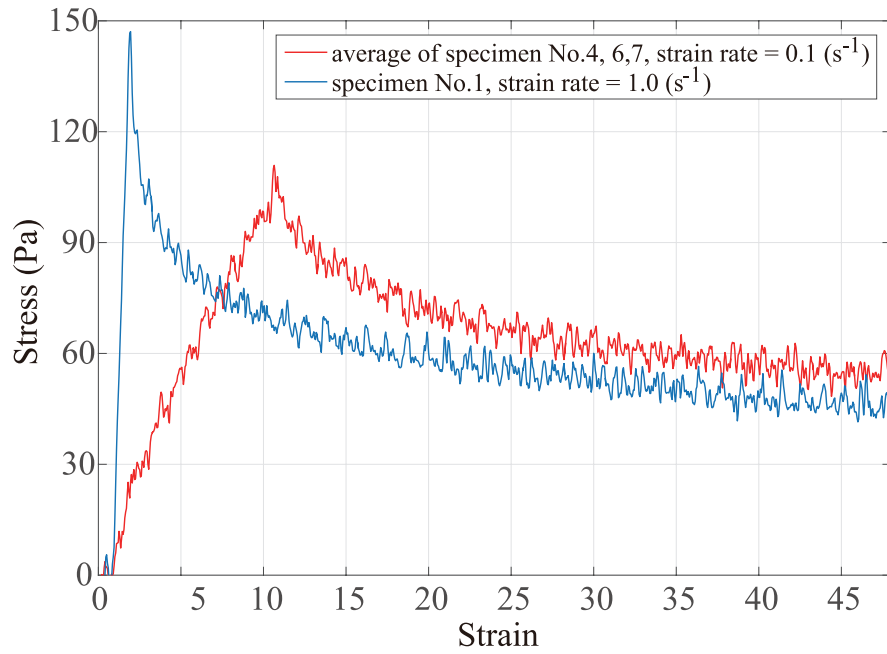


Figure 4.6: Comparison of the experimental results with strain rates of $0.01 \text{ (s}^{-1}\text{)}$ and $0.1 \text{ (s}^{-1}\text{)}$.

rate of $0.01 \text{ (s}^{-1}\text{)}$. Engineering strain and stress were calculated using the unloaded length and cross-sectional area (Eq. (3.1) and (3.2)), respectively. The noise of the force sensor was reduced using the Savitzky–Golay filter (frame size = 101).

4.4.2 Data analysis

Stress–time curves were converted to $E(t)$ curves. $E(t)$ is calculated by

$$E(t) = \frac{\text{engineering stress}}{\text{engineering strain}}. \quad (4.24)$$

An example of $E(t)$ curves is shown in Fig. 4.4 by using the experimental data of Fig. 4.3. As shown in Fig. 4.4, the $E(t)$ curves of different strains are rarely different. In other words, $E(t)$ curves are independent of strains. Therefore, when strain=0 to 0.1, the tensile and stress relaxation characteristics of brain parenchyma were characterized.

Fig. 4.5 shows the stress response of three specimens when strains change from 0 to 0.1 with a strain rate of $0.01 \text{ (s}^{-1}\text{)}$. To reduce variability in experimental results due to the individuality of specimens, the average stress response (red line in Fig. 4.5) of three specimens was calculated. The average value was smoothed using a moving average filter, and the span is 30 (blue line in Fig. 4.5). The original sampling rate of force data was reduced from 1000 to 20 (Hz).

The average stress–time curve with strain rate of $0.01 \text{ (s}^{-1}\text{)}$ and the result of one specimen with strain rate of $0.1 \text{ (s}^{-1}\text{)}$ are plotted in Fig. 4.6. As shown, the stress response of porcine brain parenchyma demonstrates strain–rate dependency. The strain rate dependency is the characteristic of viscoelastic materials.

4.5 Stress Relaxation Simulation [85]

4.5.1 Simulation scenario

The dynamic simulation was conducted after the stress relaxation tests. The simulation settings are described as follows.

- Simulation model:

The sizes of specimen models are the average sizes of specimens. The diameter is 5 (mm), and the length is 10 (mm).

- Contact nodes:

Nodes of upper and bottom surfaces ($z = 15$ and $z = 0$).

- Forced displacement of contact nodes:

Displacement of upper surface nodes is

$$\mathbf{u}_c = [0, 0, 0]^T.$$

Displacement of bottom surface nodes is $[0, 0, d]^T$. When tensile strain rate is $0.1 \text{ (s}^{-1}\text{)}$, d is

$$\begin{cases} 0 \leq t < 1 & d = 0.0 \text{ (mm)} \\ 1 \leq t \leq 11 & d = 0.1(t - 1) \text{ (mm)} \\ 11 < t \leq 88 & d = 1.0 \text{ (mm)} \end{cases} \quad (4.25)$$

When tensile strain rate is $1.0 \text{ (s}^{-1}\text{)}$, d is

$$\begin{cases} 0 \leq t < 1 & d = 0.0 \text{ (mm)} \\ 1 \leq t \leq 2 & d = 1.0(t - 1) \text{ (mm)} \\ 2 < t \leq 47 & d = 1.0 \text{ (mm)} \end{cases} \quad (4.26)$$

4.5.2 Parameter identification

Parameter N (the number of Maxwell model elements in Eq. (4.1)) was defined. The experimental data of the average stress–time curves at a strain rate of $0.01 \text{ (s}^{-1}\text{)}$ is assumed that begins at 0 (s). Therefore, it can be expressed as

$$\sigma(t) = \sigma_0 + \sum_{i=1}^N \sigma_i e^{-t/\tau_i}, \quad (4.27)$$

where σ_0 , σ_i , and τ_i were identified through curve fitting by using MATLAB Curve Fitting Toolbox™. The fitted curves are plotted in Fig. 4.7, and the coefficients of determination (R^2) are listed in Table 4.1. As shown in Fig. 4.7 and Table 4.1, the accuracy of fitted curves is not significantly different among $N = 2 - 5$. It is desirable to use a minimum number of elements in terms of evaluation and analysis. Therefore,

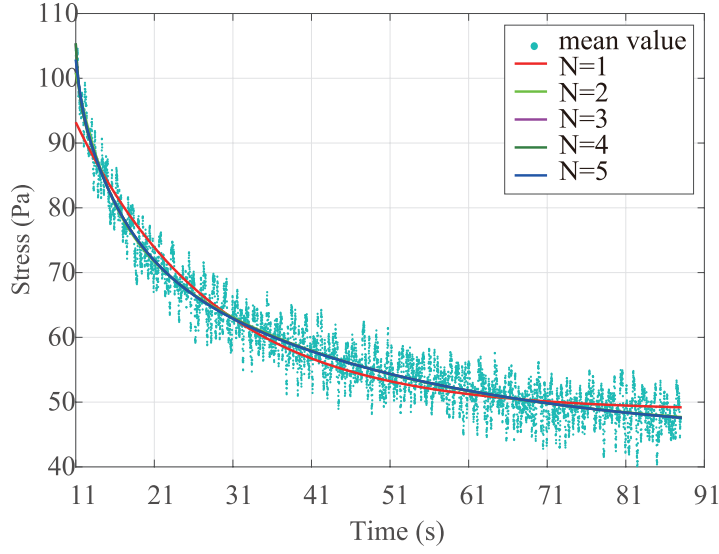


Table 4.1: Accuracy of fitted curves.

N	R^2
1	0.935
2	0.949
3	
4	
5	

Figure 4.7: Fitted curves ($N=1 - 5$) of the average stress–time curve at a strain rate of $0.01 \text{ (s}^{-1}\text{)}$.

in this paper, there are two sets of Maxwell model elements ($N = 2$) in the generalized Maxwell model.

The other parameters of generalized Maxwell model were identified by solving an optimization problem. The *fmincon()* function of MATLAB Optimization Toolbox™ was used to solve the optimization problem. As $N = 2$, the parameter vector is

$$\mathbf{x} = \begin{bmatrix} E_0 & E_1 & E_2 & \tau_1 & \tau_2 \end{bmatrix}. \quad (4.28)$$

The objective is to minimize the error between simulation result and experimental data. The strain and stress of experimental data are shown in Fig. 4.8. The stress–time curve is the moving average value shown in Fig. 4.5 with a tensile strain rate of $0.01 \text{ (s}^{-1}\text{)}$. The optimization problem is specified as follows.

- Objective function

$$\text{minimize } \mathcal{F}(\mathbf{x})$$

$$\mathcal{F}(\mathbf{x}) = \sum_t (\sigma_{\text{fit}}(t) - \sigma_{\text{sim}}(t))^2.$$

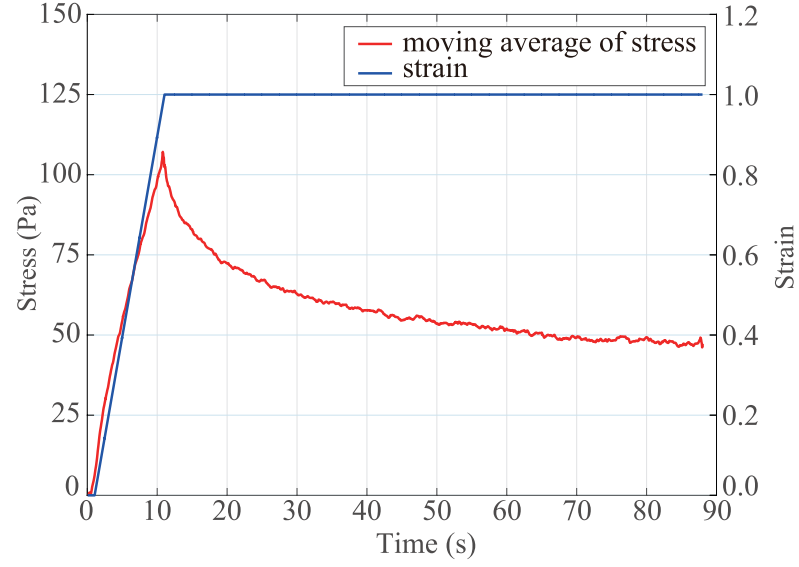


Figure 4.8: Strain provided to the simulation model and the target of stress in optimization problem (moving average in Fig. 4.5)

- Subject to

$$E_i \in [10, 1000], \quad i = 0, 1, 2$$

$$\tau_i \in [1, 50], \quad i = 1, 2.$$

- Initial parameters

$$\mathbf{x}_0 = [400 \quad 400 \quad 400 \quad 10 \quad 10].$$

- Optimized parameters

$$\mathbf{x}_{\text{opt}} = [400 \quad 411 \quad 391 \quad 18.8 \quad 4.0].$$

4.5.3 Stress relaxation simulation

Stress relaxation simulations were performed at strain rates of $0.01 \text{ (s}^{-1}\text{)}$ and $0.1 \text{ (s}^{-1}\text{)}$. Mechanical properties and viscoelastic parameters used in simulations are listed in Table 4.2. Stress–time curves obtained through simulations and experiments at strain rates of $0.01 \text{ (s}^{-1}\text{)}$ and $0.1 \text{ (s}^{-1}\text{)}$ are plotted in Figs. 4.9. Sequences of the stress relaxation simulation at the strain rate of $0.01 \text{ (s}^{-1}\text{)}$ are illustrated in Fig. 4.10.

Table 4.2: Mechanical properties and parameters of generalized Maxwell model for stress relaxation simulation

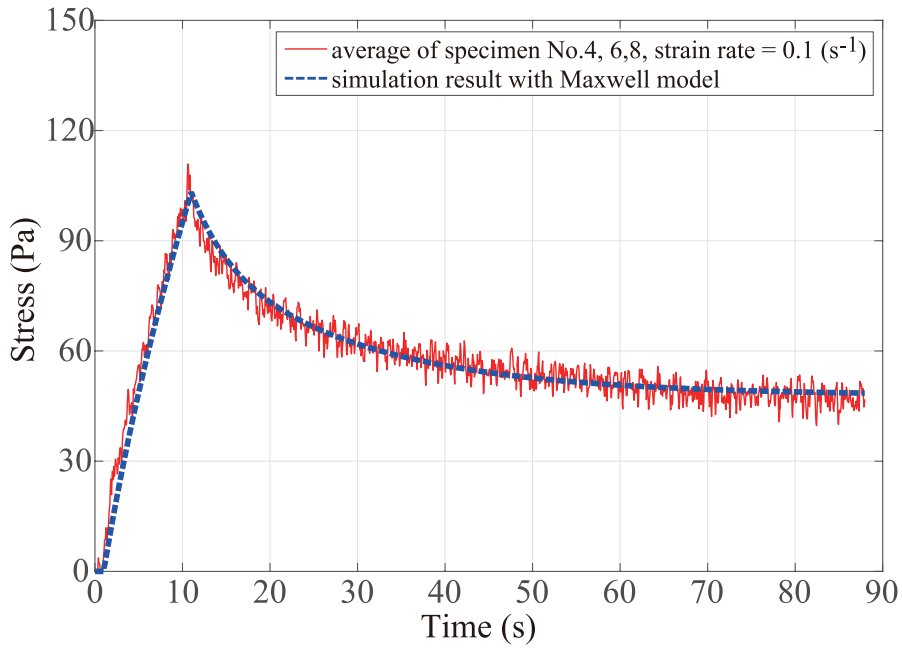
Parameters	
Number of nodes	936
Number of tetrahedral elements	4056
Sampling time	0.05 (s)
Density	1036 (kg/m ³)
Poisson's ratio	0.4925
Maxwell model scarlars	$E_0=400$ (Pa), $E_1=411$ (Pa), $E_2=391$ (Pa), $\tau_1=18.8$ (s), $\tau_2=4.0$ (s)

As the viscoelastic parameters were identified through the experimental data at the strain rate of $0.01 \text{ (s}^{-1}\text{)}$, simulation result satisfactorily reproduces the stress relaxation of the experimental result. Moreover, the strain–rate dependency is reproduced by comparing Figs. 4.9 (a) and (b).

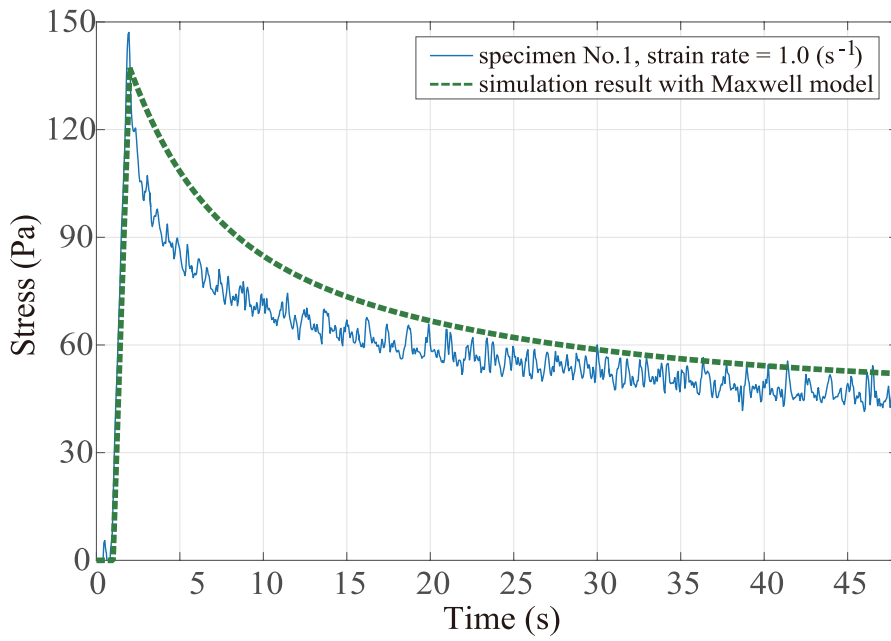
Fig. 4.9 (b) shows the comparison of the simulation result and experimental data, where slight differences are observed between simulation and experimental results. However, we consider that the error has little effect on the accuracy of the brain retraction simulation. The reasons are as follows. “stress” and “stress $\times$ time” are considered as factors that damage the nerve. The maximum stress is well represented in Fig. 4.9 (b), so that the index of “stress” is clear. The range of the difference is in several seconds. In neurosurgery, the retraction is usually continued for approximately 30 minutes. As shown in Fig. 4.11, when the scale of time in Fig. 4.9 (b) changes to long span such as 30 minutes, the difference in Fig. 4.9 (b) becomes insignificant. Therefore, the index

of “stress $\times$ time” is almost represented.

Thus, although parameters were identified using experimental results at the strain rate of $0.01 \text{ (s}^{-1}\text{)}$, the simulation satisfactorily reproduces the stress response at the strain rate of $0.1 \text{ (s}^{-1}\text{)}$. Therefore, the effectiveness of the proposed stress relaxation model to express viscoelasticity of brain parenchyma is validated.



(a) Tensile strain rate $\dot{\epsilon} = 0.01 \text{ (s}^{-1}\text{)}$.



(b) Tensile strain rate $\dot{\epsilon} = 0.1 \text{ (s}^{-1}\text{)}$.

Figure 4.9: Stress–time curve of stress relaxation simulation.

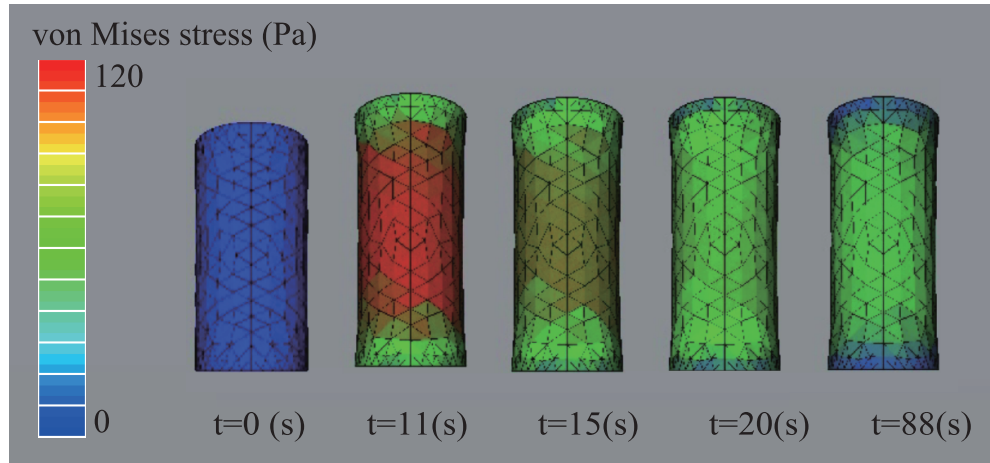


Figure 4.10: Stress relaxation simulation with identified parameters.

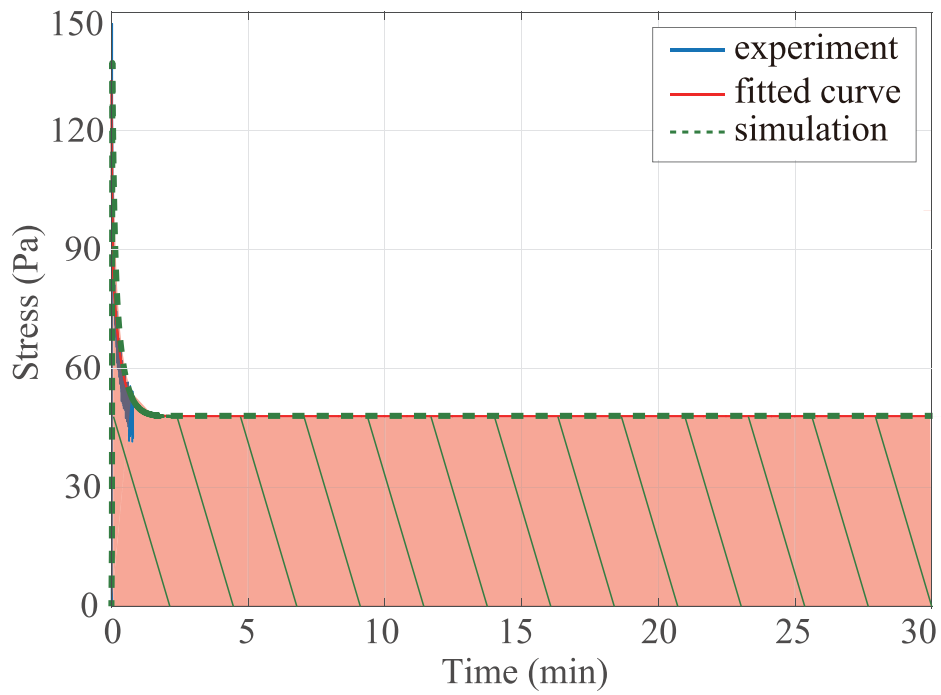


Figure 4.11: Changing the scale of time in Fig. 4.9 (b) to long span.

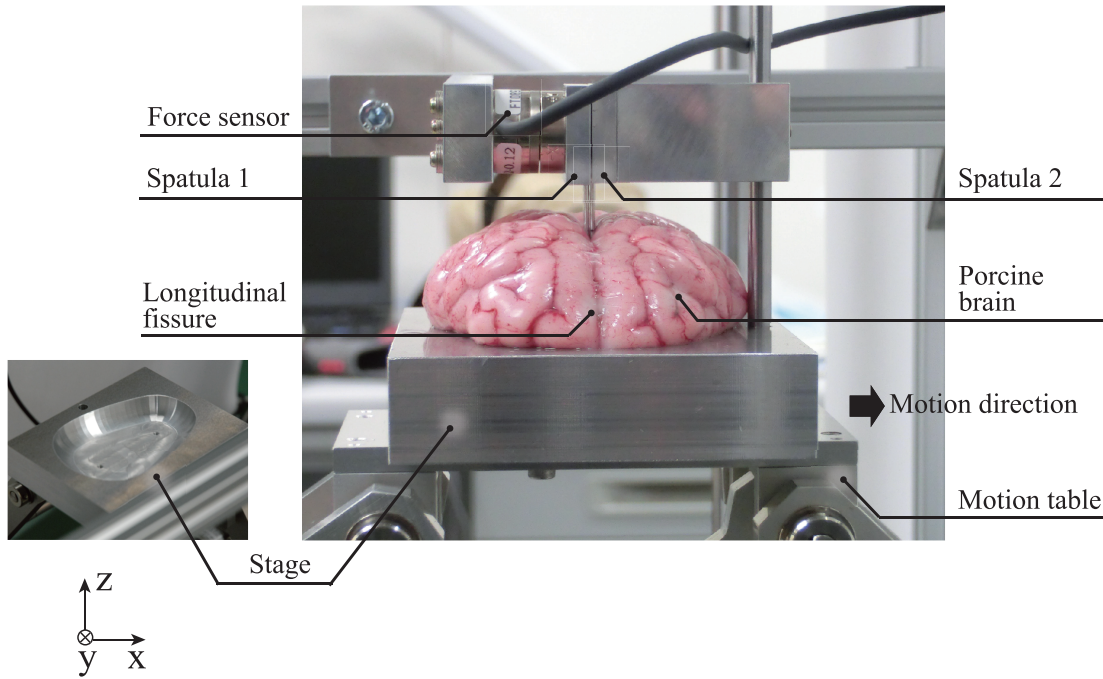


Figure 4.12: Experimental equipment for splitting brain fissure.

4.6 Retraction Experiments at Longitudinal Fissures of Cerebrum

In Section 4.5.3, the proposed stress relaxation model is verified to represent the is-coelasticity of brain parenchyma. However, the parameters of the generalized Maxwell model identified in Section 4.5.2 cannot be used for retraction simulation by using a whole cerebrum. This could be due to two reasons. First, in Section 4.4, specimens were taken from various areas of porcine brains and were mixed with white and gray matters. The experimental result in Section 4.4 showed the average value of specimens and may be not applicable to the whole brain. The other reason is that in the retraction, the influence of the arachnoid and pia mater cannot be ignored. Therefore, retraction experiments using a real cerebrum were conducted.

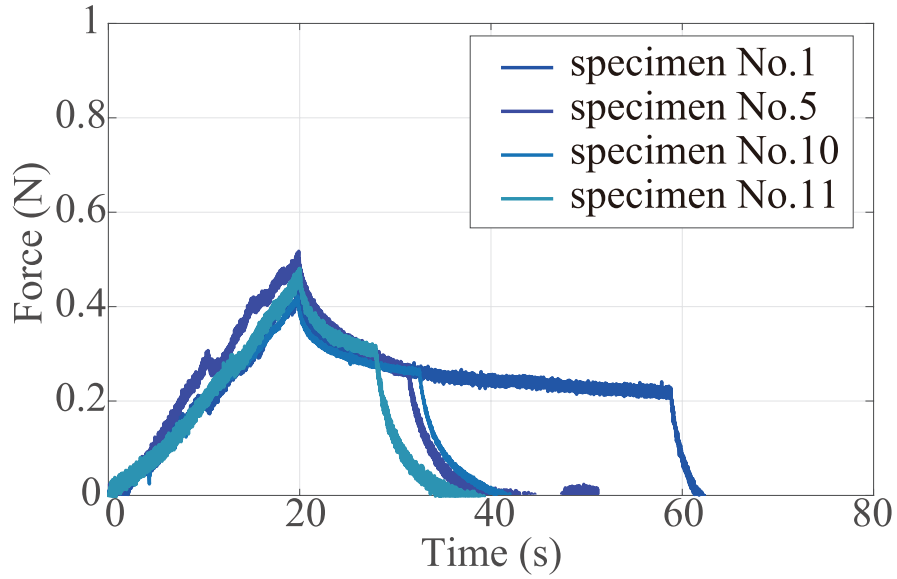
Twelve porcine heads were collected from a local slaughter house on the morning of the experiment day. Brains were harvested from heads except the dura mater. To use the brains as close to the living state as possible, brains were stored in saline at room temperature and tested in three hours after being harvested.

The experimental equipment is illustrated in Fig. 4.12. A shape of the brain bottom

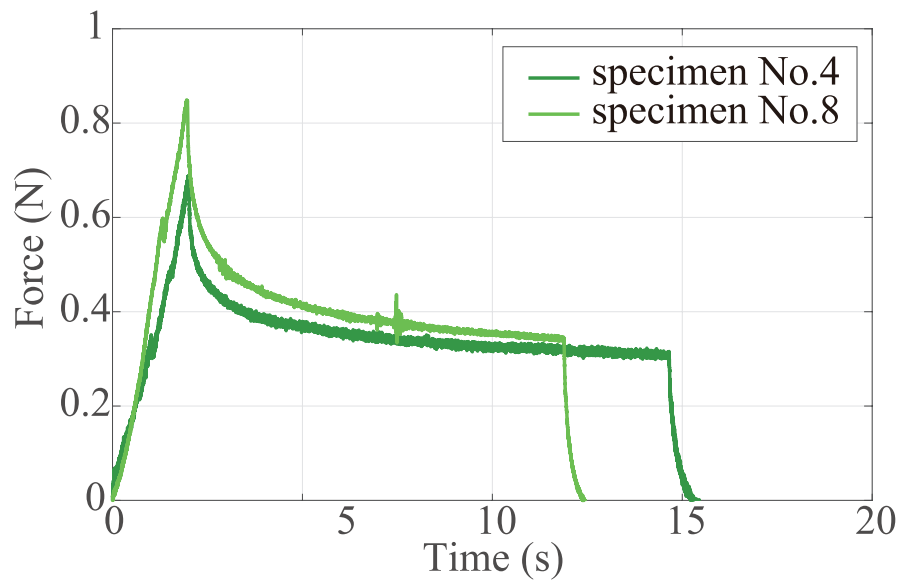
was carved into the stage and the brain was directly placed on the carved crater. In the middle of the longitudinal cerebral fissure, two spatulas were inserted, which were rectangular thin plates with thickness of 1 (mm). The contact face between spatulas and brain tissues are 5 (mm) $\times$ 15 (mm) (width of spatula $\times$ inserted depth). Spatula1 was fixed to the force sensor, and Spatula2 was fixed to the stage. The stage was in conjunction with the motion table below it, and moved with the motion table toward the positive x -axis in the experiments. As Spatula1 did not move in the experiments, it provided forced displacement ($-x$ -axis direction) on the brain. The reaction force to Spatula 1 was measured using a 6-axis force sensor named Nano 17 (ATI Industrial Automation, Inc.).

In the experiments, the velocities of the motion table were 1 (mm/s) and 10 (mm/s). When the displacement of the motion table reached 20 (mm), it remained constant to maintain the deformation for several seconds, then it returned to its initial position. Four specimens were split with velocity of 1 (mm/s) and other two specimens were split with velocity of 10 (mm/s). The stationary periods of the motion table after retraction were differentiated in each experiment. The experimental results of retraction velocity of 1 (mm/s) and 10 (mm/s) are shown in Fig. 4.13. In addition, Fig. 4.14 shows the fissure opening of an experiment with 1 (mm/s) retraction velocity.

To compare the results with different retraction velocities, experimental data were averaged at each moving speed. The averaged results are shown in Fig. 4.15. The peak forces of two retraction velocities are arranged at the same position in Fig. 4.15. The stationary periods were unified to 7.95 (s). Moreover, Fig. 4.15 shows the strain rate dependency.



(a) 1.0 (mm/s).



(b) 10 (mm/s).

Figure 4.13: Retraction force results of experiments.

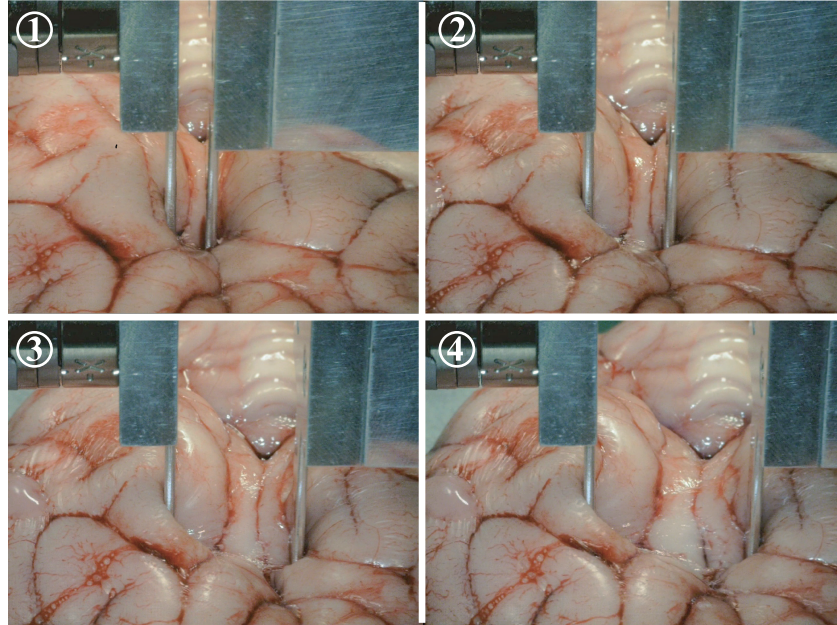


Figure 4.14: Sequences of an experiment (specimen No.11). The retraction velocity is 1 (mm/s)

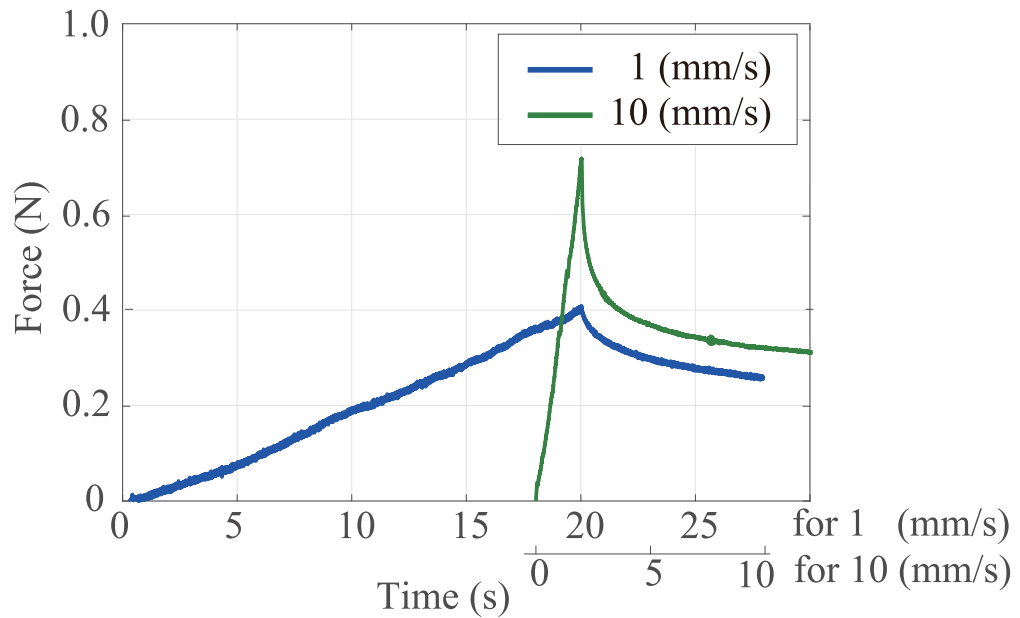


Figure 4.15: Comparison of average of experimental results of force with retraction velocity of 1 (mm/s) and 10 (mm/s)

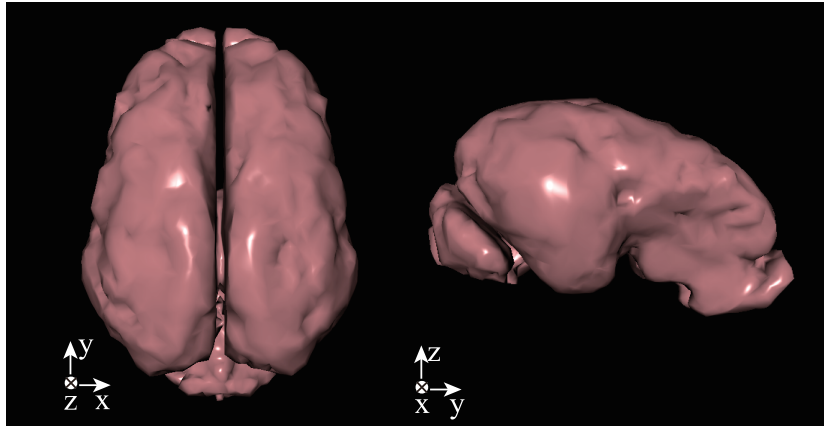


Figure 4.16: FE model of porcine brain (transformed from digital atlas [87, 56]).

Table 4.3: Size of the FE model of porcine brain.

	maximum	minimum
x -axis (mm)	27.37	-27.38
y -axis (mm)	49.36	-29.15
z -axis (mm)	30.60	-15.21

4.7 Brain Retraction Simulation

4.7.1 Simulation scenario

Saikali *et al.* constructed digital atlas of a porcine brain extracted from a six-month old female pig using MRI images [87]. The atlas data is publicly available on the web page [56]. The porcine heads used in our experiments were also almost six-months old. Moreover, size and weight of the brains used in [87] and specimens of our experiments were similar. Therefore, the digital atlas can be used as a simulation model to represent the experiments. The condition that differs is that there is membrane and some connective tissues of brains used in experiments but there are no such tissues in the digital atlas. In this study, when the fissure was retracted, the effect of membrane and connective tissues was not significant hypothetically.

The sequence of the retraction experiment was faithfully reproduced in the simulation with the following settings.

- Simulation model:

The digital atlas of a porcine brain obtained from [56] were transformed to an FE model. As the digital atlas are built from MRI images, there are some cavities inside and fragments outside. Such improper parts were modified when the digital atlas were transformed to an FE model. The shape of the FE model is shown in Fig. 4.16 and the sizes are listed in Table 4.3. The brain is considered as homogeneous.

- Contact nodes:

The surface nodes of the brain model in contact with two spatulas and a stage were classified into contact nodes. The position of nodes touching Spatula 1 is $\mathbf{p}_{sp1} = \begin{bmatrix} x & y & z \end{bmatrix}$, where the surface nodes are in the area

$$-1 < x < 0, 15 < y < 20, z_{max} < z < z_{max} - 15,$$

where z_{max} is the nodes in this area with maximum z . Similarly, the position of nodes touching Spatula 2 is $\mathbf{p}_{sp2}$, with the surface nodes in the area:

$$0 < x < 1, 15 < y < 20, z_{max} < z < z_{max} - 15.$$

The depth of the carved part of stage is 10 (mm). The position of nodes contacting the stage meet surface nodes in the area $z_{min} < z < z_{min} + 10$.

- Forced displacement of contact nodes:

In the experiments, Spatula 2 and the stage were moved, while Spatula 1 was fixed (see Fig. 4.12). Hence, in the simulation, displacement of nodes in contact with Spatula 1 is

$$\mathbf{u}_{csp1} = \begin{bmatrix} 0 & 0 & 0 \end{bmatrix}^T.$$

Displacements of nodes in contact with Spatula 2 and the stage are

$$\mathbf{u}_{csp2} = \mathbf{u}_{cstage} = \begin{cases} \begin{bmatrix} 1.0 \cdot t & 0 & 0 \end{bmatrix}^T & \text{for } 1.0 \text{ (mm/s)} \\ \begin{bmatrix} 10 \cdot t & 0 & 0 \end{bmatrix}^T & \text{for } 10 \text{ (mm/s)} \end{cases}.$$

The final displacement of x -axis of $\mathbf{u}_{csp2}$ and $\mathbf{u}_{cstage}$ was 20 (mm).

- Reaction force:

The number of nodes in contact with Spatula 1 was counted as N_c in Eq. (2.41) to calculate the reaction force.

4.7.2 Parameter identification

Here, a five elements generalized Maxwell model ($N = 2$ in Eq. (4.3)) was used for the analysis.

Firstly, E_0 was identified. In the stationary condition, $\ddot{\mathbf{u}} = \mathbf{0}$, $\dot{\mathbf{u}} = \mathbf{0}$, the reaction force $\mathbf{f}_{\text{re}}(t)$ converges to $\mathbf{f}_{\text{re}} = \mathbf{K}_{\text{re}}\mathbf{u}_{\text{re}}$ when $t \rightarrow \infty$, where subscript 're' expresses that the terms are related with nodes that contact with Spatula 1. $\mathbf{K}_{\text{re}}$ is the stiffness matrix calculated using E_0 . Hence E_0 can be identified by the convergence value of the reaction force $\mathbf{f}_{\text{re}}$. The convergence value $\mathbf{f}_{\text{re}}(\infty)$ was identified by curve fitting using *fitype()* of MATLAB®. The targets of fitting are force-time curve plotted in Fig. 4.15 from 2 (s) to 12 (s) of 10 (mm/s). According to the results of fitting, the convergence values of experimental data with retraction velocity of 10 (mm/s) was 0.3282 (N). The coefficients of determination (R^2) was 0.9953. Finally, E_0 was identified to be 3224.0 (Pa).

Parameters of the generalized Maxwell model are identified by solving an optimization problem. The *patternsearch()* function of MATLAB Global Optimization Toolbox™ was chosen to solve the optimization problem. The initial guess is calculated by the parameters of the fitted curve. The conditions and result of the optimization problem are listed as follows.

- Parameter vector:

$$\mathbf{x} = \begin{bmatrix} E_0 & E_1 & E_2 & \tau_1 & \tau_2 \end{bmatrix}$$

- Objective function:

$$\text{minimize } \mathcal{F}(\mathbf{x})$$

$$\mathcal{F}(\mathbf{x}) = \sum_{t=0 \text{ (s)}}^{t=12 \text{ (s)}} (f_{\text{fit}}(t) - f_{\text{sim}}(t))^2,$$

where $f_{\text{fit}}(t)$ is the average force result of the experiments with retraction velocity of 10 (mm/s) (green line in Fig. 4.15)

- subject to

$$\begin{aligned} E_i &\in [0, 10^4], \quad i = 1, 2 \\ \tau_i &\in [10^{-2}, 10], \quad i = 1, 2 \end{aligned}$$

- Initial guess:

$$\mathbf{x}_0 = \begin{bmatrix} 1696.5 & 1950.9 & 2.61 & 0.20 \end{bmatrix}$$

- Optimization result:

$$\mathbf{x}_{\text{opt}} = \begin{bmatrix} 3488.5 & 6718.9 & 2.29 & 0.23 \end{bmatrix}$$

4.7.3 Simulation results

Table 4.17 lists the simulation conditions and identified parameters of the proposed model for retraction simulation. In addition, the sequences of the simulation with 10 (mm/s) retraction velocity are shown in Fig. 4.18. The reaction forces of the simulation with retraction velocity of 1.0 (mm/s) and 10 (mm/s) are plotted in Fig. 4.19.

As shown in Figs. 4.19 (a) and (b), by the comparison of the simulation result and experimental data, experimental force is well represented. The results of simulation can be used to avoid nerve damage during retraction.

4.8 Summary

The details of stress relaxation model are presented in this chapter. The stress relaxation model is based on the generalized Maxwell model while it is added an inertial term. By adding the inertial term, computational calculation is more stable even when the stiffness matrix becomes singular.

Stress relaxation tests were conducted using cylindrical specimens taken from porcine brain. Properties of porcine brain for the proposed model were identified, and the usefulness of the proposed model was verified.

Moreover, brain retraction experiments were conducted using porcine brains. Next, an FE model of a porcine brain was built. By using the FE model, parameters of the proposed model were identified and brain retraction simulations were performed. The comparison of the simulation and experimental results proved that the proposed model can simulate brain retraction.

Figure 4.17: Simulation conditions and identified parameters

Parameters	
Number of nodes	5212
Number of tetrahedral elements	25102
Sampling time	0.05 (s)
Density	1036 (kg/m ³)
Poisson's ratio	0.4925
Identified parameters	$E_0=3224.0$ (Pa), $E_1=3488.5$ (Pa), $E_2=6718.9$ (Pa), $\tau_1=2.29$ (s) $\tau_2=0.23$ (s)

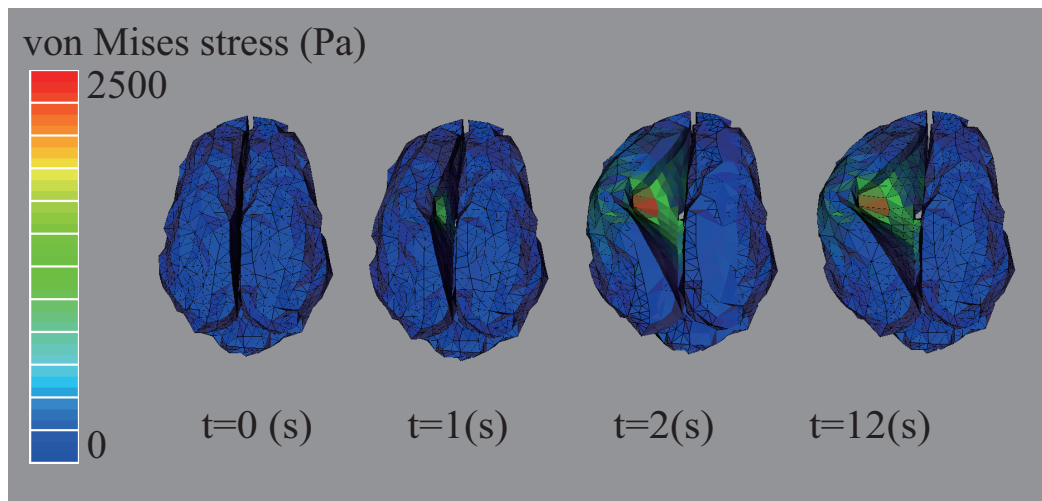
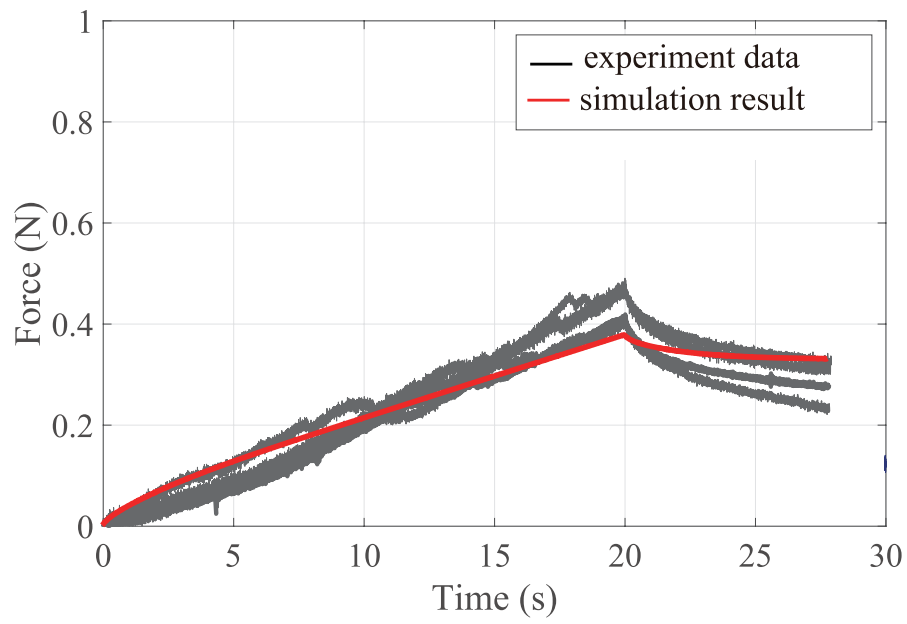
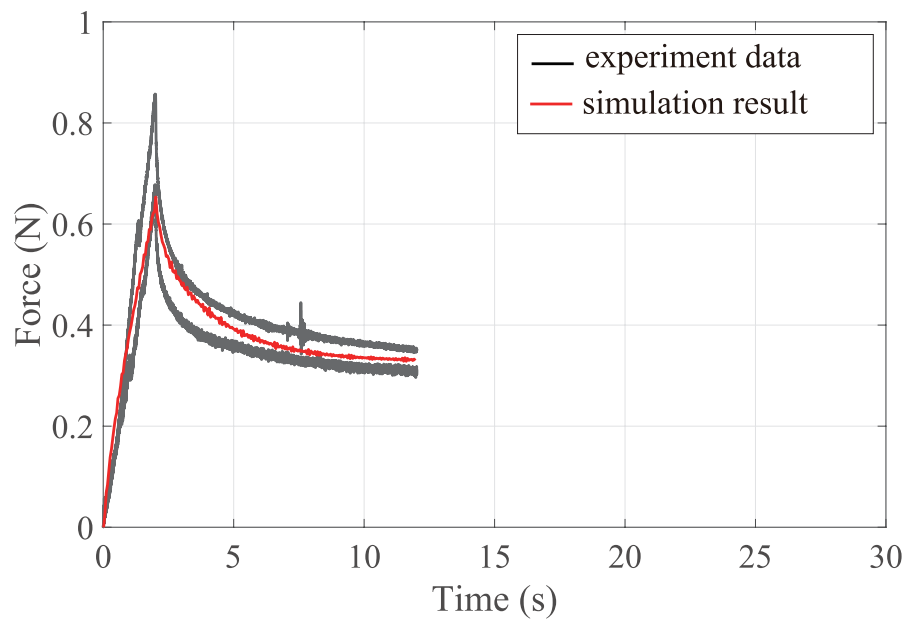


Figure 4.18: Retraction simulation



(a) 1.0 (mm/s).



(b) 10 (mm/s).

Figure 4.19: Retraction simulation result and experimental data.

Chapter 5. Conclusion

5.1 Summary

In this thesis, numerical models of blunt dissection and brain retraction for a neurosurgery simulation are described.

Damage fracture models

Brain parenchyma has strong ductility behavior and strain—rate dependence. Few studies on the damage of brain tissue at low strain rates have been conducted. In this study, tensile tests were performed using porcine brain tissues. Mechanical properties of the damage of brain tissues were measured.

Ductile fracture is generally expressed through damage models based on continuum damage mechanics. The existing damaged models are applicable for ductile fracture of metals, and cannot reproduce the ductile behavior of brain parenchyma. Two damage fracture models were proposed in this study.

Tensile simulations were performed using the damage fracture models, and the parameter of these models were identified. It was confirmed that the damage fracture models can accurately reproduce the damage behavior of the brain parenchyma.

Stress relaxation model

A stress relaxation model was constructed to simulate viscoelastic properties of the brain tissues. The model considers the inertia term and it improves numerical calculation to make them more stable.

Columnar specimens were extracted from porcine brains. Stress relaxation experiments were conducted. In addition, the parameters of the stress relaxation model were identified. Stress relaxation simulations were conducted. Simulation results showed that

the proposed stress relaxation model can reproduce stress relaxation of brain parenchyma.

Moreover, retraction experiments were performed using a whole porcine brains. The mechanical properties of the porcine brain when it is retracted at longitudinal fissure of cerebrum were measured. Parameters of the whole brain by using the model were identified. Retraction simulations were conducted using an FE model of whole porcine brain. Simulation results reproduced the stress of the brain retraction at longitudinal fissure of cerebrum.

5.2 Future Work

In the future, the following work is expected to be completed.

Introducing models to a neurosurgery simulation

Damage fracture and stress relaxation models will be implemented in a neurosurgery simulation. The simulation can be used for brain tumor removal. Moreover, the simulation will be evaluated by surgeons. Based on the evaluation, the system and calculation algorithm will be improved. Finally, a neurosurgery simulation which can faithfully simulate actual neurosurgery operations and be used for training and preoperative planning will be developed.

Development of surgery support system

In neurosurgeries, the head of the patient is fixed before the surgery, and alignment with CT or MRI images taken before surgery is performed. Such a surgery support system is called neuronavigator. By combining the neuronavigator and the developed neurosurgery simulation, a more safety surgery support system will be constructed. For example, the position of the surgical tools is detected using the optical position detector. The contact between the surgical tools and the brain tissues is judged from the position of the head registered in the neuronavigator. The expected stress can be presented visually, which helps to prevent unnecessary brain damage.

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